GAIT & CLINICAL

GCMAS

MOVEMENT ANALYSIS SOCIETY

22nd Annual Meeting
Salt Lake City, UT
May 22-26, 2017

TABLE OF CONTENTS

WEDNESDAY, 24 May 2017

THURSDAY, 25 May 2017

FRIDAY, 26 May 2017

E POSTERS

GCMAS 2017 Program

PROGRAM – AT – A -GLANCE

Time	Mon, May 22 2017	Tues, May 23 2017		Wed, May 24 2017		Thurs, May 25 2017		Fri, May 26 2017
		Registration open 7am - 5pm		Registration open 7am - 5pm		Registration open 7am - 5pm		Registration 7-10 am
7:00 AM				Dual Fluoroscopy and Gait - Dr. Anderson		Breakfast Session - CMLA		Breakfast Session - GCMAS Committees
7:30 AM								
7:45 AM								
8:00 AM	Finewire EMG Course	Finewire EMG Advanced Intrepretation and Practice Course		Podium Session 1 Cerebral Palsy I	Exhibitors Open	Podium Session 4 Pathologic Gait/ Prostheses	Exhibitors Open	Podium Session 7 Cerebral Palsy III
8:30 AM								
9:00 AM								
9:30 AM				Exhibitor/Poster Session 1		Exhibitor/Poster Session 3		Break
10:00 AM								Podium Session 8 Data Methods/ Modeling
10:30 AM								
11:00 AM				Keynote Speaker: Todd Carver, MS		Keynote Speaker: Mike Aiona, MD		
11:30 AM								Awards
12:00 PM				Vendor Luncheon Session in Vendor Area		GCMAS Business Lunch OR Vendor Lunch		
12:30 PM								
1:00 PM		Tutorial #1 Role of Gait Analysis in CMT	Tutorial #2 Selective Control in CP					
1:30 PM				Podium Session 2 Sports		Podium Session 5 Pediatric Gait/ Cerebral Palsy II		
2:00 PM	Registration Open 2-4 pm							
2:30 PM								
3:00 PM		Break		Exhibitor/Poster Session 2		Exhibitor/Poster Session 4		
3:30 PM		Tutorial #3 Successful Practices for Gait Analysis	Student Symposium					
4:00 PM				Podium Session 3 Foot/Ankle		Podium Session 6 Trunk and Upper Extremity		
4:30 PM							Exhibitors Breakdown	
5:00 PM								
5:30 PM								
6:00 PM		Welcome Reception				GCMAS Banquet		
6:30 PM								
7:00 PM				User Group Meetings				
7:30 PM								
8:00 PM								
8:30 PM								

To Access E-POSTERS, Visit OPEN CONFERENCE for GCMAS 2017: https://openconf.org/GCMAS2017

GCMAS President Welcome Letter

Dear Colleagues:

It's with great pleasure that I welcome you to the 22nd annual meeting of the Gait and Clinical Movement Analysis Society in Salt Lake City! Our Conference Co-Chairs, Bruce MacWilliams and Bo Foreman, have worked hard to pull together all the nitty gritty details that will make this meeting a great success. I hope each of you will take a moment this week to express your thanks to Bruce and Bo.

Your Program Chairs for this meeting, Mark McMulkin and Dr. Jason Rhodes, have sorted through all your abstract submissions, managed the review and scoring process and constructed a strong scientific program. They have boldly met the challenge of new-to-GCMAS technology for this aspect of planning for our annual meeting, using OpenConf to manage the submission and review of abstracts. This online resource was also instrumental in building the formal proceedings, and should provide you with easy access to abstracts throughout the meeting. If you have any feedback on these aspects of the 2017 experience, positive or negative, please share them with a GCMAS Board member or with Mark or Jason. We are already working on the 2018 annual meeting and will be discussing whether to continue with OpenConf, so your feedback would be most welcome as a factor in that decision.

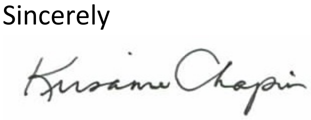
I'd also like to acknowledge and welcome our Exhibitor attendees. We genuinely appreciate the continued support from vendors for the GCMAS and our annual meeting, and I hope each one has a productive time in Salt Lake City, meeting and greeting new and established customers. Attendees should have ample time during the meeting to take advantage of the Exhibit booths and talk with these expert representatives about their latest products.

We are a unique society, bringing together scientists and physicians, therapists and engineers, to better understand human movement. Whether you are a long-time member or brand new to this society, don't hesitate to participate in the Q&A or engage speakers informally after a session. Make the most of your "downtime" to network with potential new colleagues and catch up with old friends. Be prepared to be invigorated and energized by new knowledge or new collaborations. And finally, please remember that the Society only moves forward with the help of members who volunteer time, energy and service to one of the Councils. If you would like to contribute, we have a council for you!

As I will be passing the baton of GCMAS President to Jason Long at this meeting, I would like to thank the membership for the honor of serving in this role for the past two years. It has been a pleasure to represent you and your interests and I look forward to continued involvement with the Society as past-president.

Thank you for coming, thank you for participating and please enjoy Salt Lake City!

Sincerely



Krisanne Chapin, PhD
GCMAS President, 2015-2017

GCMAS Program Chairs Welcome Letter

Dear Colleagues,

We are excited to have you join us in Salt Lake City for the 2017 Gait and Clinical Movement Analysis Society Meeting. You are part of a group of scientists, clinicians, students, and researchers with diverse backgrounds yet all connected by an interest in the analyses of movement and posture. We have developed a scientific program that speaks to your diversity and your commitment to advancing our profession. Additionally, we hope that this meeting will provide a chance to catch up with old friends and converse with new colleagues.

We have introduced a couple new ideas for this GCMAS conference. First, the entire program complete with abstract accessibility is available online (https://www.openconf.org/GCMAS2017/openconf.php). Second, we have adopted e-Posters for this conference. e-Posters are accessible online from any internet accessible device throughout the conference. Please view these e-Posters and engage the authors during the open times in the schedule. Finally, CME and CEU credits are available for the scientific content of the program so complete declaration of credits and evaluations to receive your credits.

We have incorporated a broad Podium, Poster, and e-Poster Scientific Program that includes sessions on Cerebral Palsy, Foot/Ankle, Pathologic Gait, Upper Extremity, Data Methods/Modeling, and Sports. In addition, there are tutorials targeted on clinical and technical topics of interest to attendees. Keynote speakers will address clinical aspects of the Gait and Clinical Movement Analysis Society as well as provide insight into the analysis of cycling.

We encourage you to take advantage of the diverse scientific, educational, and creative topics. Make a point of visiting the posters and engaging the authors in discussion. During or after the podium sessions, ask a question or make a comment to the presenter. During the poster sessions, lunches, and other open times be sure visit the exhibitors to learn about the most current technologies and techniques.

In closing, we want to acknowledge the individuals who submitted abstracts, our reviewers, the moderators and the presenters. We also want to acknowledge the hard work of the conference committee, in particular our local hosts Bo Foreman and Bruce MacWilliams; Jenna Yentes and her committee for selecting such excellent tutorials; Jenny Kent for organizing the student activities.

The GCMAS annual meeting has been designed to invigorate your clinical and research practices by exposing you to the latest thinking on topics in movement analysis. We hope that you enjoy the conference, the beautiful Salt Lake City area.

Mark McMulkin and Jason Rhodes
GCMAS 2017 Conference Program Co-Chairs

GCMAS Conference Chairs Welcome Letter

Dear Colleagues,

On behalf of the University of Utah and Shriners Hospitals for Children, we welcome you to Salt Lake City, Utah for the 22nd annual Gait and Clinical Movement Analysis Society meeting. We would like to thank Mark McMulkin, PhD and Jason Rhodes, MD for their time, effort and dedication to organizing an excellent scientific program. In addition, we extend a warm welcome to our keynote speakers, Todd Carver, MS, Co-Founder of Retül University and Head of Human Performance for Specialized Bicycles and Michael Aiona, MD, Chief of Staff and Director of the Movement Analysis Laboratory at Shriners Hospitals for Children – Portland. We would also like to thank Andy Anderson, PhD, for sharing his expertise on dual fluoroscopy and gait during the breakfast session. Additional thanks to Nick Van Stratt, Technology Services Director, Helms Briscoe for his tireless attention to the endless issues of this meeting; Carrie Grant, from the University Guest House who is coordinating this event and is responsible for organization all the activities, providing the food, and setting up at the meeting and exhibitor space. Thanks also to GCMAS President, Krisanne Chapin, PhD and GCMAS Administrator Sahar Hassani, MS for their insight and guidance; Jenna Yentes, PhD for organizing the student symposium; Carole Tucker, PT PhD and Sherry Backus, PT DPT MA for organizing the pre-meeting EMG course; as well as all the podium and poster presenters, for their contributions to the success of this meeting. Last, but not least, we would like to express our gratitude to the vendors for informing attendees of the latest advances in equipment, measurement, and utilization; supporting student education through travel funds to the society, and ensuring the financial success of this meeting.

We are hoping the weather is warm and dry for those of you with the time to explore the various opportunities in the Salt Lake City area. These include visiting our beautiful foothills that are a short walk from the conference and mountain canyons (Emigration, Millcreek, Big Cottonwood, and Little Cottonwood) that are a short drive away. In addition, within a few mile radius we encourage you to visit Red Butte Garden, This is The Place Heritage Park, Hogle Zoo, Natural History Museum, Fort Douglas Military Museum, Utah Museum of Fine Arts to name a few local attractions. Within the city, you can visit Temple Square and enjoy shopping downtown at the City Creek Center. Runners and hikers can find many opportunities to explore the University of Utah campus and foothills and enjoy views of the city and local wildlife. Please feel free to seek us out if you have any concerns or questions during the meeting and we hope you enjoy your stay in Salt Lake City.

Sincerely,

Bo Foreman, PT, PhD and Bruce MacWilliams, PhD
GCMAS 2017 Conference Co-Chairs

General Information

Target Audience and Purpose

Attendees of GCMAS meetings are professionals from diverse disciplines who are actively involved in human movement analysis and research in clinics, academic labs, and other settings.

The 2017 Meeting brings these individuals together to:

- Advance scientific knowledge, technical capabilities, and clinical practice in the field of human movement.
- Provide a forum for professional interaction and collaboration among physicians, allied health professionals, engineers and scientists that will facilitate improved care for individuals with movement disorders.
- Educate and encourage the career development of students interested in human movement research.

Learning Objectives

The 2017 GCMAS Meeting will help participants to:

1. Describe and evaluate technologies for measuring kinematic, kinetic, EMG data, in-shoe pressure profiles, and other biomechanical variables of interest.
2. Interpret and critically compare kinematic, kinetic, and EMG data from patients with movement disorders, both before and after treatment.
3. Summarize current “best practices” for treating gait abnormalities in persons with cerebral palsy, stroke, other movement disorders and musculoskeletal disorders.
4. Define common clinical and technical terms used by physicians, therapists and engineers who evaluate or treat persons with movement disorders.

Badges

Please wear your badges to all conference events; your badges are required for admission to the tutorials, podium and poster sessions, ancillary workshops, and social events.

Cellular Phones

Out of respect for your fellow attendees, please set your phones to silent mode and answer all calls outside the podium, poster, and exhibitor areas.

Updates on Facebook and Twitter

Updates on upcoming conferences and Society activities are posted on Facebook (www.facebook.com/gcmas) and Twitter (www.twitter.com/gcmas). Follow along!

Student and Trainee Information

Student participation is an integral part of the GCMAS annual meeting. We enthusiastically invite students and postdoctoral trainees to take advantage of the following opportunities.

Student Symposium

On Tuesday May 23rd the GCMAS will host its sixth annual Student Symposium. The student symposium will focus on addressing topics directly affecting graduate students, post-docs, and trainees. This session will be free of charge to all attendees and will be immediately followed by the Welcome Reception.

The symposium will consist of stations set up for round table discussion. Moderators at each table will be experienced mentors familiar with the topic being explored. These mentors will begin with a very short presentation to highlight important points, then attendees will be invited to ask any questions. The emphasis will be on discussion and the sharing of experience and advice. This will also be a time to get to know other students in GCMAS.

Student Mentorship Program

Since 2009, the GCMAS Education Committee has organized a mentorship program in which student or postdoctoral trainee mentees are matched with a volunteer mentor based on mutual research interests and the student's desired career path. The goal of this program is to facilitate dialog among students and professionals in a supportive manner, where knowledge, skills, and experiences are shared, and where networking relationships are cultivated. Typically, mentors and mentees meet over lunch or during a scheduled break at the annual meeting. To participate in this program, you must submit your information for matching early in the year. If you are currently participating in the program, make sure that you take the time to meet one-on-one with your mentor. Be sure to ask your mentor questions regarding their insight into finding jobs, hiring processes and even research tips. If you would like to participate in the event in the future, please contact the student representative to the GCMAS Board of Directors, Jenny Kent, from the University of Nebraska at Omaha at jkent@unomaha.edu.

Student Travel Scholarship Program

Travel scholarships will again be available to student and postdoctoral trainee members of GCMAS who present their research at the annual meeting. This program is made possible through the generosity of the GCMAS membership and our sponsors. Applications are reviewed by the GCMAS Education Council prior to the Annual Meeting. Recipients are selected based on their demonstrated potential to advance scientific knowledge, technical capabilities, and/or clinical practice in the field of human movement. Be sure to check the conference website each year for the application deadline.

Student Travel Sponsors

The Gait and Clinical Movement Analysis Society is grateful to the following companies for their sponsorship of the Student Travel Scholarship Program. Scholarship recipients were selected by the Education Council based on their demonstrated potential to advance scientific knowledge, technical capabilities, and/or clinical practice in the field of human movement.

Venue

University Guest House and Conference Center

The University Guest House and Conference Center is located on the University of Utah campus in the beautiful historic section of Fort Douglas on the east side of the Salt Lake valley. The hotel and conference center were the home for the 2002 Winter Olympic athletic village and the beautiful Fort Douglas area is a very peaceful and scenic area of campus. Being located adjacent to the Salt Lake City TRAX line makes a trip to downtown Salt Lake just minutes away.

The University Guest House is part of the Foothill Cultural District of Salt Lake City. Some of Salt Lake's biggest attractions are within walking distance of the Guest House or, at most, a short bus ride away. Enjoy a walk through extensive gardens, visit prestigious museums or the Hogle Zoo.

GCMAS Councils and Board of Directors

Executive Officers

President: Krisanne Chapin, PhD
President-Elect: Jason Long, PhD
Past-President: Bruce MacWilliams, PhD
Secretary-Treasurer: Kirsten Tulchin-Francis, PhD

Council Chairs

2017 Conference Council: Bo Foreman, PhD, Bruce MacWilliams, PhD
2017 Program Council: Mark McMulkin, PhD, Jason Rhodes, MD
Awards Council: Aviva Wolf, PhD
Communications Council: Braden Romer, PhD
Education Council: Jenna Yentes, PhD
Membership Council: Ross Chafetz, DPT, PhD
Reimbursement Council: Sylvia Öunpuu, MSc
Research Council: Jinsup Young, PhD
Standards Council: Howard J. Hillstrom, PhD
Student Representative: Jenny Kent

Advisory Board

Past-President: Frank Chang, MD
Past-President: Tom Novacheck, MD
Member at Large: Howard Hillstrom, PhD

Administration

Society Administrator: Sahar Hassani, MS

GCMAS Board Meetings

WEDNESDAY, 24 May 2017

TIME	EVENT	ROOM
7:00A - 8:00A	GCMAS Board Meeting	Alpine

THURSDAY, 25 May 2017

TIME	EVENT	ROOM
7:00A - 8:00A	GCMAS Board Meeting	Alpine
12:00P-1:30P	GCMAS Business Lunch	Douglas Ballroom

FRIDAY, 26 May 2017

TIME	EVENT	ROOM
7:00A-8:00A	GCMAS Board Meeting	Alpine

List of Moderators

The program council graciously thanks the following individuals for their contributions as session moderators and discussion leaders at the GCMAS 2017 Meeting.

Glen Baird, MD

John Henley, PhD

Susan Kanai, PT

Stephen Hill, PhD

Roy Davis, PhD

Howard Hillstrom, PhD

Marilynn Wyatt, PT

Audrey Zucker-Levine, PhD

Sherry Backus, PT

Frank Chang, MD

Joanna Roybal, PT

Carole Tucker, PhD

Susan Rethlefson, PT

Kristan Pierz, MD

Adam Rozumalski, PhD

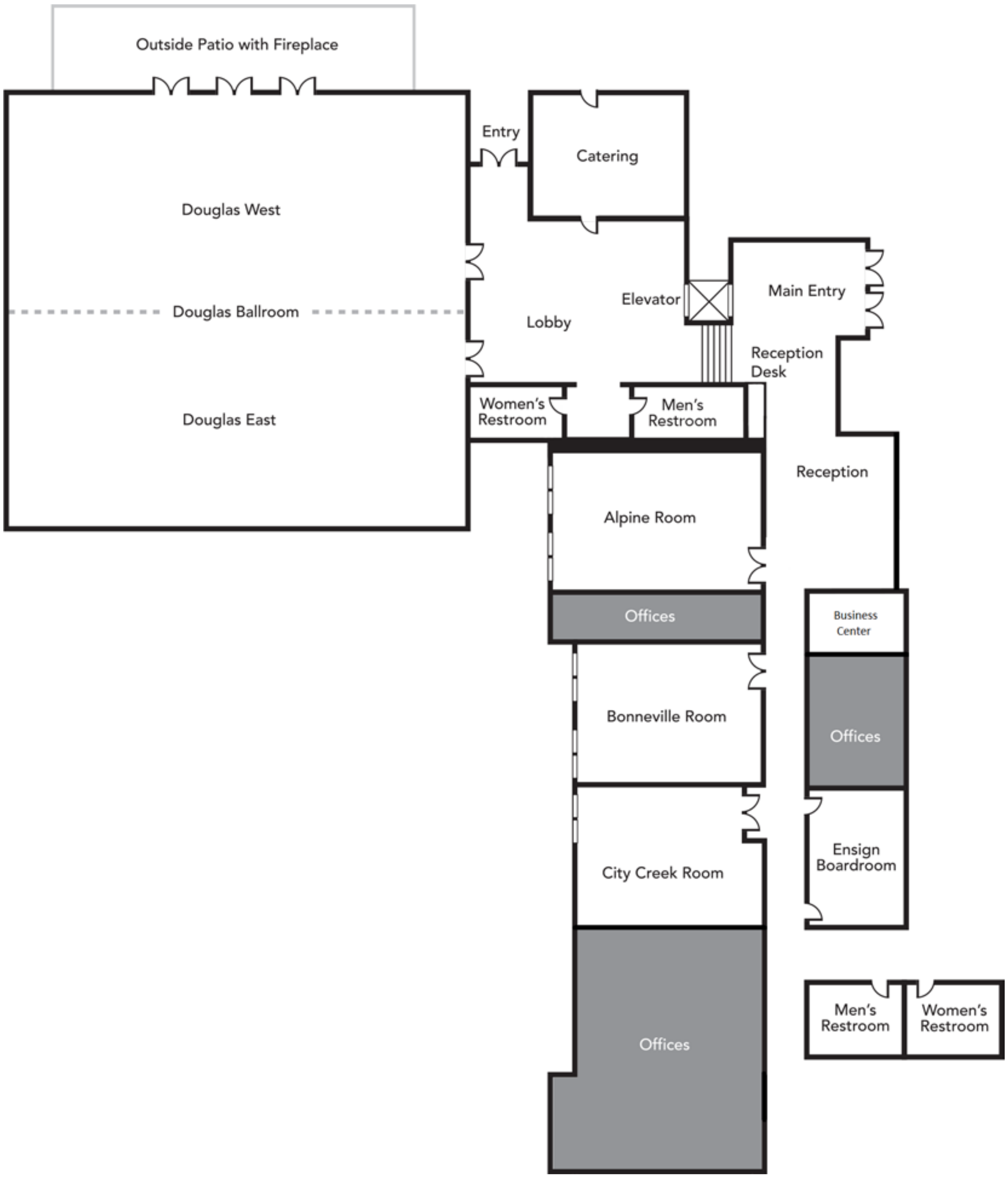
Jason Long, PhD

List of Reviewers

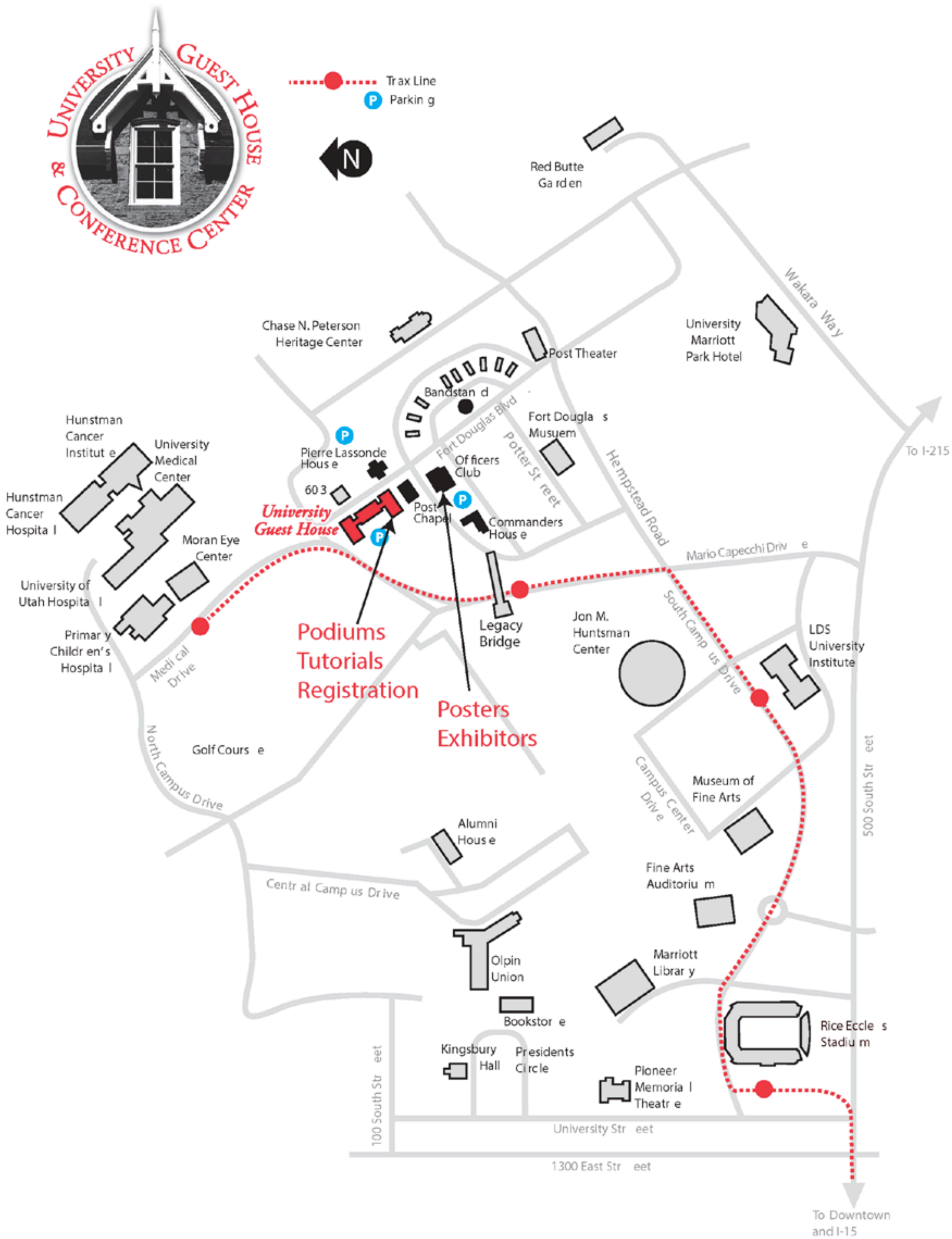
We would like to thank the following individuals for serving as
Reviewers for the 2017 GCMAS Scientific Program

Michael Aiona
Gordon Alderink
Allison Arnold
Sherry Backus
Glen Baird
Christina Bickley
Alec Black
Amy Bodkin
James Carollo
Li-Shan Chou
Stephen Cobb
Krista Coleman Wood
Kevin Cooney
Roy Davis
Sandi Dennis
Jing Feng
Bo Foreman
Adam Fullenkamp
George Gorton
Adam Graf
Chris Hass
Steven Irby
Kelly Jeans
Cathy Johnson
Susan Kanai
Kenton Kaufman
Robert Kay
William Ledoux
Jason Long
Bruce MacWilliams
Matthew Major
Dennis Matthews
Jean McCrory
Tom Novacheck
Donna Oeffinger
Michael Orendurff
Sylvia Öunpuu
Kyria Petuskey
Stephen Piazza
Kristan Pierz
Susan Rethlefsen
Jim Richards
Braden Romer
Adam Rozumalski
Michael Schwartz
Amy Shuckra
Susan Sienko
Kachun Siu
Jean Stout
Pamela Thomason
Carole Tucker
Kirsten Tulchin-Francis
David Westberry
Hank White
Audrey Zucker-Levin

University Guest House Floor Plan



University Guest House and Officers Club



List of Exhibitors

AMTI
176 Waltham St
Watertown, MA 02472 USA
617.926.6700
www.amti.biz

APDM, Inc
2828 Southwest Corbett Ave, Suite 130
Portland, OR 97201 USA
503.446.4055
www.apdm.com/mobility

Bertec Corporation
6171 Huntley Rd, Suite J
Columbus, OH 43229 USA
614.543.1127
www.bertec.com

C-Motion, Inc
20030 Century Blvd, Suite 104A
Germantown, MD 20874 USA
301.540.5611
www.c-motion.com

CIR Systems Inc. / GAITRite
12 Cork Hill Road, BLDG #2
Franklin, NJ 07416 USA
888.482.2362
www.gaitrite.com

Cometa Systems
Cometa srl
Bareggio, Italy
+39.340.987.0881
www.cometasystems.com

Delsys, Inc
23 Strathmore Rd
Natick, MA 01760 USA
508.545.8200
www.delsys.com

List of Exhibitors (continued)

Easy Walking
1478 Dillon Road
Maple Glen, PA 19002 USA
215.654.1626
www.easy-walking.com

Idoneus Digital
186-8120 No. 2 Road Suite #243
Vancouver, BC V7C 5J8 Canada
604.368.5959
idoneusdigital.com

Kistler Instruments
30280 Hudson Drive
Novi, MI 48377 USA
248.668.6900
www.Kistler.com

Motion Analysis Corporation
3617 Westwind Blvd
Santa Rosa, CA 95403 USA
707.579.6500
www.motionanalysis.com

Motion Lab Systems, Inc
15045 Old Hammond Highway
Baton Rouge, LA 70816 USA
225.272.7364
www.motion-labs.com

Noraxon USA, Inc
15770 N Greeway-Hayden Loop #100
Scottsdale, AZ 85260 USA
800.364.8985
www.noraxon.com

OptiTrack
3658 SW Deschutes Street
Corvallis, OR 97333 USA
541.207.7976
www.optitrack.com

List of Exhibitors (continued)

ProtoKinetics, LLC

60 Garlor Dr
Havertown, PA 19083 USA
610.449.4879
www.protokinetics.com

Qualysis North America, Inc

1630 Old Deerfield Rd, Suite 206
Highland Park, IL 60035 USA
847.945.1411
www.qualisys.com

Tekscan, Inc

307 W First St
South Boston, MA 02127-1309 USA
617.464.4500
www.tekscan.com

Vicon

7388 S Revere Parkway, Suite 901
Centennial, CO 80112 USA
303.799.8686
www.vicon.com

Conference Materials

Thanks to the following for providing handout materials:

Department of Orthopaedics

Exhibitor Booth Locations

Officers Club

	EXHIBITOR		EXHIBITOR
	Premium		**Deluxe**
2	CIR Systems/GAITRite	11	Easy Walking, Inc.
3	BERTEC Corp.	12	ProtoKinetics, LLC
4	AMTI, Inc.	13	APDM, Inc.
5	Motion Analysis Corp.	14	Motion Lab Systems, Inc.
6	OptiTrak	15-16	Tekscan, Inc.
	Deluxe		**East Room**
7	Delsys, Inc.	17	Idoneus Digital
8	C-Motion, Inc.	18	Kistler Instruments
9	Noraxon USA, Inc.	19	Cometa Systems
10	Vicon	20-21	Qualysis North America, Inc.

Daily Schedule: Monday 22 May 2017

Pre-Conference Workshop

Kinesiological Electromyography: Fine-Wire Techniques

University of Utah Department of Physical Therapy

8:00A-7:00P

TIME	TOPIC & ACTIVITY
8:00A – 8:30A	Course Introduction & Logistics and Course Pre-test
8:30A – 9:00A	General Guidelines and Techniques for Kinesiological Fine-Wire EMG
9:00A – 9:15A	Fine-wire Insertion Techniques & Demonstration: EDL/TA*
9:15A – 12:00P	Anatomy Lab
12:15P – 12:45P	Lecture/Discussion: Data Interpretation with Boxed Lunch
12:45P – 1:45P	Practice Session 1: EDL/TA*
1:45P – 2:15P	Lecture/Discussion: Instrumentation & Quality Assessment
2:15P– 3:30P	Practice Session 2: Fine-Wire Insertion Technique: Tib Post/ FHL
3:30P - 3:45P	Break
3:45P – 4:45P	Practice Session 3: Fine-Wire Insertion Technique: Thigh (Rectus/VL/TFL/Glut Medius)
4:45P – 5:15P	Lecture/Discussion: Basic Interpretation
5:15P – 5:45P	Fine-wire Insertion Techniques & Demonstration: Shoulder & Scapula, Proximal Hip (Hip Flexors)
5:45P – 6:45P	Practice Session 4: Fine-Wire Insertion Muscles TBD
6:45P – 7:00P	Final Discussion & Questions and Course Post-test

TIME	TOPIC & ACTIVITY
2:00P – 4:00P	Registration

Pre-Conference Workshop

Kinesiological Electromyography: Fine-Wire Techniques

Monday, May 22nd
8:00A-7:00P

Course Faculty:

Sara Mulroy PT, PhD:	*Director, Pathokinesiology Laboratory, Rancho Los Amigos, National Rehabilitation Center*
Sylvia Öunpuu, MS:	*Director of Research, Center for Motion Analysis, Connecticut Children's Medical Center*
Jean Stout, PT, MS:	*Research Physical Therapist, Gillette Children's Specialty Healthcare*
Marilynn Wyatt, PT, MA:	*Director, Gait Analysis Laboratory, Naval Medical Center San Diego*
Sherry Backus, PT, DPT, MA:	*Clinical Supervisor, Leon Root, MD Motion Analysis Laboratory, Hospital for Special Surgery*
Carole Tucker, PT, PhD, PCS:	*Associate Professor of Physical Therapy & Electrical Engineering, Temple University*

Description: This hands-on workshop will provide clinicians with the opportunity to learn and perform fine-wire EMG insertions. Course content will include didactic material covering the following topics: 1) gross anatomy, neuroanatomy, nerve and muscle physiology foundational concepts; 2) Electromyography – concepts; 3)pertinent anatomy and physiology fine-wire EMG general guidelines (indications, contraindication, consent, and techniques); 4) EMG instrumentation, choice of equipment, data processing and process quality assessment; 5) Hazards and complications, instrumentation and troubleshooting; and 6) Preparation, patient informed consent for procedure, and post-test care, and 7) clinical interpretation. Each participant will have the opportunity to practice fine-wire insertions for 3 – 4 muscles in small groups under the guidance of experienced kinesiological fine-wire EMG instructors.

Learning Objectives: Upon completion of this training program, participants will:

- Demonstrate understanding of the indications, precautions and contraindications, risks, post-test care involved in performing kinesiological fine-wire electromyographic studies.
- Demonstrate understanding of gross anatomy and neuroanatomy, muscle and nerve physiology,
- Demonstrate an understanding of the instrumentation, choice of instruments, hazards and complications, troubleshooting used for kinesiological EMG
- Perform fine-wire EMG insertion techniques safely and effectively for selected muscles.

Daily Schedule: Tuesday 23 May 2017

Pre-Conference Workshop

Fine-Wire EMG: Advanced Interpretation and Practice

University of Utah Department of Physical Therapy

8:00A-12:00P

TIME	TOPIC & ACTIVITY
8:00A – 8:15A	Course Introduction & Logistics and Course Pre-test
8:15A – 8:45A	EMG data collection – instrumentation and signal quality testing,
8:45A – 9:15A	New hardware and software approaches
9:15A – 10:00A	EMG Data Advanced Interpretation Case Discussions 1
10:00A – 10:15A	Break
10:15A – 11:30A	EMG Data Advanced Interpretation Case Discussions 2
11:30A – 11:45A	Alumni: Tips from the Field
11:45A – 12:00P	Final Discussion & Questions and Course Post-test

GCMAS 2017 Annual Meeting

University of Utah Guest House

12:00P-5:30P

TIME	EVENT	
7:00A – 5:00P	Registration	
1:00P – 3:00P	TUTORIAL 1: Role of Gait Analysis in CMT Sylvia Öunpuu, MSc Kristan Pierz, MD	TUTORIAL 2: Selective Control in CP Marcia B. Greenberg MS, PT, KEMG Loretta A. Staudt MS, PT Eileen Fowler PhD, PT
3:00P – 3:30P	Break	
3:30P – 5:30P	TUTORIAL 3: Successful Practices for Gait Analysis James Carollo PhD, PE John Henley PhD Tom Novacheck MD Susan Rethlefsen DPT	STUDENT SYMPOSIUM
6:00P – 7:30P	WELCOME RECEPTION Red Butte Garden 300 Wakara Way, Salt Lake City, UT 84108 Shuttles and Walking Maps Available	

Pre-Conference Workshop

Fine-Wire EMG: Advanced Interpretation and Practice

Tuesday, May 23rd
8:00A-12:00P

Course Faculty:

Sara Mulroy PT, PhD:	*Director, Pathokinesiology Laboratory, Rancho Los Amigos, National Rehabilitation Center*
Sylvia Öunpuu, MS:	*Director of Research, Center for Motion Analysis, Connecticut Children's Medical Center*
Jean Stout, PT, MS:	*Research Physical Therapist, Gillette Children's Specialty Healthcare*
Marilynn Wyatt, PT, MA:	*Director, Gait Analysis Laboratory, Naval Medical Center San Diego*
Sherry Backus, PT, DPT, MA:	*Clinical Supervisor, Leon Root, MD Motion Analysis Laboratory, Hospital for Special Surgery*
Carole Tucker, PT, PhD, PCS:	*Associate Professor of Physical Therapy & Electrical Engineering, Temple University*

Description: This 4-hour course will provide individuals experienced in fine-wire EMG an opportunity to participate in lectures and case based discussions to improve their skills in problem-solving EMG data quality issues, and to improve their understanding and skills in interpreting clinical EMG data. This course consists predominantly of case based discussion and short lectures. Participants should have attended the full-day FWEMG course this year, or in the past, or have equivalent content knowledge and experience. If a participant has limited fine wire insertion experience, the full-day course is the more appropriate course. If you have any questions, please email the course directors.

Learning Objectives: Upon completion of this training program, participants will:

- Be able to identify poor quality EMG data (surface primarily – principles apply to fine wire)
- Be able to interpret EMG within context of clinical cases
- Demonstrate an understanding of the instrumentation, choice of instruments, hazards and complications, troubleshooting used for kinesiological EMG
- Describe the emerging EMG instrumentation and analyses
- Perform fine-wire EMG insertion techniques safely and effectively for selected muscles.

Tutorial #1

Role of Gait Analysis in CMT

Tuesday, May 23rd
1:00-3:00P

Instructors: Sylvia Öunpuu, MSc and Kristan Pierz, MD

Purpose: The purpose of this course is to describe the clinical impairments for patients with Charcot-Marie-Tooth (CMT) and the associated gait findings to build a basis for determining the most appropriate treatment for gait issues in this patient group. A comprehensive motion analysis protocol for the assessment of persons with CMT will also be reviewed.

Audience: This course is for physicians, mid-level practitioners, physical therapists, orthotists, kinesiologists and engineers who are interested in a more detailed understanding of CMT in the context of motion analysis outcomes.

Prerequisite Knowledge: Participants should have a minimum of basic level skills in gait analysis data interpretation including joint kinematics and kinetics.

Abstract: CMT represents a spectrum of neurological dysfunction and is also characterized by progressive decline in strength. Treatment must be determined based on the individual patient's specific presentation in terms of impairment and associated gait issues. Individuals with CMT frequently present with foot and ankle problems such as pain, weakness, deformity, and difficulty with shoe wear. These issues will be discussed in the context of clinical impairments typical to persons with CMT. Gait analysis is a useful tool for documenting and analyzing gait pathology and has allowed for the identification of three characteristic presentations of the foot and ankle in CMT: flail foot, cavovarus foot, and toe walking. These three foot and ankle presentations in persons with CMT will be described in detail from the clinical impairments to the gait kinematic and kinetic findings. Then the treatment options will be discussed for the particular presentation with example treatment outcomes presented. This course will provide participants with information necessary to identify gait patterns and critically evaluate treatment options. The components of a comprehensive motion analysis for this patient population will also be reviewed. The focus will be on children and youth.

Learning objectives: At the completion of this tutorial, attendees will:

1) Be able to describe CMT in terms of the etiology and associated clinical impairments
2) Be able to recognize the three different foot/ankle presentations for CMT disease
3) Be able to identify the most appropriate treatment approaches for each CMT presentation
4) Be able to understand the impact of treatment in terms of motion analysis outcomes

Tutorial #2

Selective Control in CP

Tuesday, May 23rd
1:00-3:00P

Instructors: Marcia B. Greenberg MS, PT, KEMG, Loretta A. Staudt MS, PT, Eileen Fowler PhD, PT

Purpose: To instruct experienced clinicians in the use and administration of a standardized clinical tool for assessment of selective voluntary motor control (SVMC) of the lower extremity in patients with spastic cerebral palsy (CP) using SCALE (Selective Control Assessment of the Lower Extremity). The reliability and validity of the SCALE has been established (Fowler et. al. Dev Med Child Neurol 51:607-614, 2009). The tool and its clinical and research applications will be presented. Participants will have an opportunity to practice scoring patients using videotaped assessments. The use of this tool in clinical decision-making will be discussed.

Audience: Clinicians evaluating patients with CP and individuals conducting research on the clinical or functional characteristics of children and adults with CP.

Prerequisite Knowledge: A basic understanding of gait kinematics and CP

Abstract: The role of SVMC assessment is often over-looked in the treatment planning of patients with spastic CP. In this tutorial, the role of SVMC in clinical practice and research will be discussed. SCALE administration, including patient positioning, examiner instructions and score sheet, will be explained. The criteria for each SVMC grade will be described, providing participants with the knowledge and skill to independently assess SVMC. Participants will have an opportunity to use SCALE to assess all lower extremity joints on a variety of videotaped patient examples exhibiting a range of SVMC. Discussion and feedback on the participants' skills will be provided. The relationship of SVMC to other impairments such as strength and spasticity will be presented and the use of SCALE scores in surgical decision making and its role in therapeutic treatment interventions will be discussed.

Learning objectives: At the completion of this tutorial, attendees will:

1) Describe the purpose, content and administration of SCALE for evaluation of SVMC.
2) Describe the scoring system for SCALE
3) Describe the role of SVMC assessment in clinical decision-making and evidence-based practice.
4) Describe the research supporting the relationship between SVMC and various biomechanical and clinical factors in patients with spastic CP

Tutorial #3

Successful Practices for Instrumented Gait Analysis: Insight from CMLA Accreditation

Tuesday, May 23rd
3:30-5:30P

Instructors: James Carollo PhD, PE, John Henley PhD, Tom Novacheck MD, Susan Rethlefsen DPT

Purpose: The concept of this tutorial is to present laboratory procedures and successful practices culled from 8 years of independent clinical motion laboratory accreditation reviews, by members of and applicants to the Commission for Motion Laboratory Accreditation (CMLA). By sharing these lessons learned, the general community of clinical motion analysis practitioners can be better equipped to provide the highest quality services to their clients, and be prepared for achieving Full Accreditation by CMLA.

Audience: 1) Individuals interested in applying for clinical motion laboratory accreditation.
2) Any professionals serving or supporting instrumented gait analysis facilities who are interested in successful practices that assure the highest laboratory quality.

Prerequisite Knowledge: None.

Abstract: The Commission for Motion Laboratory Accreditation (CMLA) is the only independent clinical motion laboratory accreditation body in the western hemisphere, and has been reviewing accreditation applications from the public since 2008. During this time theCommission has observed from their applicants a variety of ways to approach the goal of delivering high quality clinical motion analysis services. This tutorial will share information inspired by successful practices from accredited laboratories in the US, and will focus on 4 major components that comprise the unique measurements and procedures necessary to deliver quality clinical motion analysis services. These 4 areas are 3D Motion, Dynamic EMG, Physical Exam measures, and Clinical Interpretation/Recommendations. We outline each of these areas as follows:

<u>3D motion</u>: This presentation will look at three methods to test the quality and accuracy of the 3D motion and kinetic measurements made in a motion analysis laboratory. The first covers the use and application of the SAMSA device to provide a standard data set to evaluate your equipment, configuration and lab environment. The second covers the use of the CalTester which provides a tool and methodology to measure the spatial and temporal correspondence and alignment of force and motion data. The third explores a method that uses constant aspects of your 3D motion data to evaluate the accuracy and quality of the collection and processing motion data given that each data collection trial may present a unique and dynamic challenge.

<u>Dynamic EMG</u>: This presentation focuses on methods to assure accuracy, repeatability, and highest quality of electromyographic recordings used during movement analysis. The focus will be on ways to confirm proper muscle placement with surface and fine wire electrodes prior to recording, and periodic quality assurance practices to improve recording accuracy and confidence in muscle timing information. The successful practices described will be delivered in the context of how to best accomplish these procedures in a clinical environment and communicate these practices on an accreditation application.

Tutorial #3 (continued)

Physical Exam: This presentation will focus on methods of attaining initial competency, maintaining competency and measuring reliability of staff for all aspects of physical examination related to gait analysis testing. The importance of setting up standardized procedures will be emphasized. Methods will be presented for training new staff in physical examination procedures and establishing their reliability with existing staff. Methods will be presented for routine assessment of inter-and intra-examiner reliability for range of motion, spasticity, selective control and strength assessments, keeping in mind the practical aspects of assessing reliability in a busy laboratory environment. Examples will be given to illustrate all of the above, based on the faculty's personal experience going through the CMLA accreditation process.

Interpretation/Recommendations: In the assessment of individuals with walking difficulties, many pieces are needed to create a comprehensive picture of the orthopaedic and neurological impairments confronting a patient. The preceding presentations covered how to ensure the quality of these components. The final step is to accurately interpret this information to generate a comprehensive problem list and individualized treatment plans that will most benefit the patient. This presentation will cover protocols and practices a laboratory can incorporate to guarantee initial competency of new staff, ensure ongoing quality of seasoned staff and ensure that interpretation of clinical data and recommendations for treatment are of the highest quality.

Learning objectives: At the completion of this tutorial, attendees will be able to describe motion laboratory procedures and practices to provide the highest quality motion laboratory analysis for the clients seen

Student Symposium

Tuesday, May 23rd
3:30-5:30P

Organizer: Jenny Kent, MS

Panel: Edmund Cramp, Motion Lab Systems, Inc.
An industry perspective

Janet Dufek, PhD., University of Nevada, Las Vegas
An academic perspective

Adam Rozumalski, PhD., Gillette Children's Specialty Healthcare
A clinical perspective

Description: This year's student symposium features three well-established professionals from the clinical world, from academia and from industry. Each panel member will talk about their career path and share experiences and advice. The talks will be followed by invited questions and open discussion. The symposium is an informal event to encourage discussion amongst attendees and to enable new connections to be formed early on in the conference.

Daily Schedule: Wednesday 24 May 2017

GCMAS 2017 Annual Meeting

University of Utah Guest House and Officers Club

7:00A – 5:30P

TIME	TOPIC & ACTIVITY	ROOM
7:00A – 5:00P	Registration	Lobby
7:00A – 7:45A	Breakfast Session: Dual Fluoroscopy and Gait, Andy Anderson, PhD	Douglas Ball Room
8:00A – 5:00P	Exhibitors Open	Officers Club
8:00A – 9:30A	Podium Session 1 – Cerebral Palsy I	Douglas Ball Room
9:30A – 11:00A	Exhibitors / Attended Posters - Odd Numbered	Officers Club
11:00A – 12:00P	Keynote Speaker - Todd Carver, MS	Douglas Ball Room
12:00P – 1:30P	Vendor Luncheon	Officers Club
1:30P – 3:00P	Podium Session 2 – Sports	Douglas Ball Room
3:00P – 4:00P	Exhibitor / Posters	Officers Club
4:00P – 5:30P	Podium Session 3 - Foot/Ankle	Douglas Ball Room
7:00P – 9:00P	User Group Meetings	

Breakfast Session

Application of High-speed Dual Fluoroscopy to Study Human Gait

Wednesday, May 24th
7:00A - 7:45A

Presenter: Andrew E. Anderson, PhD
Associate Professor, Department of Orthopaedics
University of Utah

Abstract: This breakfast session will focus on the University of Utah's experience with the development, validation, and deployment of high-speed dual fluoroscopy to quantify in-vivo arthrokinematics of human joints. The presentation will begin with a broad overview of the technical aspects of dual fluoroscopy, with special emphasis on the importance of validation. An overview of the scientific studies conducted using the Utah dual fluoroscopy system will be discussed to provide a testament to the unique power of this technology; studies will cover the use of dual fluoroscopy to quantify the accuracy of standard optical tracking techniques as well as Utah's experience investigating clinical populations. Current and future research projects will be highlighted as well. The presentation will conclude with a discussion of special considerations concerning dual fluoroscopy, such as financial costs, integration with existing motion capture equipment, risk to human subjects, and challenges with data collection and analysis.

Learning objectives: At the completion of this tutorial, attendees will:

1. Possess an understanding of the hardware and software components of a dual fluoroscopy system
2. Understand how validation studies should be designed and executed
3. Learn how dual fluoroscopy can answer research questions that are difficult/impossible to address using standard optical motion capture techniques
4. Comprehend the nuances associated with imaging living human subjects, both from technical and regulatory points of view
5. Have a basic understanding of the cost and time investment needed to implement a dual fluoroscopy at their home institution

Podium Session #1

CEREBRAL PALSY I

MODERATED BY: **Glen Baird, MD:** Chief of Staff
Shriners Hospitals for Children, Portland, OR

John Henley, PhD: Director, Gait Analysis Laboratory
Nemours/A. I. duPont Hospital for Children, Wilmington, DE

1. **Evidence of Enhanced Moment Arms Following Patellar Tendon Advancement Surgery**
   Moria Bittmann, Rachel Lenhart, Michael Schwartz, Tom Novacheck, Darryl Thelen

2. **Comparison of Patellar Advancement Techniques in Children with Patella Alta and Cerebral Palsy**
   Alex Tagawa, Jason Rhodes, Frank Chang, James Carollo

3. **Crouch Gait Modification During a Short Walking Exercise**
   Audrey Parent, Annie Pouliot-Laforte, Pierre Marois, Laurent Ballaz

4. **Relationship Between Patella Height, Crouch Severity, and Outcomes Following Surgical Treatment**
   Moria Bittmann, Rachel Lenhart, Michael Schwartz, Tom Novacheck, Darryl Thelen

5. **Does Distal Femoral Extension Osteotomy with Patellar Tendon Advancement in Individuals with Cerebral Palsy Help in the Long-term?**
   Elizabeth Boyer, Jean Stout, Jennifer Laine, Sarah Gutknecht, Lucas Henrique Olivera, Meghan Munger, Michael Schwartz, Tom Novacheck

6. **Surgical Treatment of Pes Planovalgus in Ambulatory Children with Cerebral Palsy: Effect on Gait as Characterized by Multi-Segment Foot Motion Analysis and Foot Deformity**
   Mike Schwartz, Sue Sohrweide, Roy Wervey, Nickolas Nahm, Tom Novacheck

7. **The Periacetabular Osteotomy Improves Radiographic and Gait Functional Outcomes of Adolescents with Cerebral Palsy**
   Daniel J Sucato, Kirsten Tulchin-Francis, Adriana De La Rocha, Wilshaw Stevens Jr, David A Podeszwa

8. **Rate of Force Development in Isometric Strength Tests Are Related to Self-Reported Physical Activity in Adults with Cerebral Palsy**
   James Carollo, Meghan Colip, Patricia Heyn

Evidence of Enhanced Moment Arms Following Patellar Tendon Advancement Surgery

Moria F. Bittmann[1], Rachel L. Lenhart[1], Michael H. Schwartz[2,3], Tom F. Novacheck[2,3], Darryl G. Thelen[1]
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INTRODUCTION

Patella tendon advancement (PTA), often coupled with a distal femoral extension osteotomy (DFEO), is increasingly being used to treat crouch gait in children with cerebral palsy. The primary purpose of the PTA is to re-tension the quadriceps, which go slack with the DFEO, and also to address quadriceps insufficiency. The PTA procedure is typically performed in a way that moves the patella position from alta to baja [1]. However, it remains unclear what the long-term implications of patella baja are on the function of the knee extensor mechanism. A recent modeling study in our lab suggests that patella alta may enhance patellar tendon moment arms (PTMA) in crouched postures. In this study, we build on the prior work by modeling the PTA procedure and comparing predictions with clinical metrics. The purpose of was twofold: 1) Simulate the effects of PTA-induced changes in patella position on the PTMA, and 2) Retrospectively analyze radiographs of children who underwent PTA to test for evidence of post-surgical enhancement of moment arms.

CLINICAL SIGNIFICANCE

This study provides modeling and empirical evidence that improved knee extensor function could arise, in part, from enhanced patellar tendon moment arms after PTA. However, our model also suggests that placing the patella in a baja position may compromise moment arms in flexed postures, and thus impede flexed-knee tasks such as chair rise and stair climbing.

METHODS

Patella Height and Moment Arm Measurements: We retrospectively analyzed pre- and post-surgical radiographs of 63 limbs of 39 cerebral palsy patients (Age: 13±3 years) who underwent coupled PTA/DFEO procedures for crouch gait (average knee flexion during stance: 50±15°). Patella height was measured using the Caton-Deschamps (CD) index, and is reported as a normative (z-score) measure relative to healthy adults. To calculate the PTMA we first define the tibiofemoral flexion axis via a cylindrical fit of the femoral condyles (GCFC). The PTMA was then computed as the perpendicular distance between the GCFC and the line of action of the patellar tendon, as defined using anatomical landmarks [2].

PTA Simulation: We used a validated knee model that included cartilage contact, ligaments, and capsule restraints, all acting on six degree-of-freedom representations of the tibiofemoral (TF) and patellofemoral (PF) joints [3]. The knee was incorporated into a generic lower extremity musculoskeletal model in OpenSim (www.simtk.org). The patella insertion was translated along the tibia to simulate 8 patella heights ranging from patella alta to patella baja (Fig. 1A). For each patella position, we simulated passive knee flexion between 0 and 90° and computed the PTMA throughout the motion.

Post-Operative Changes in Moment Arms: For each subject, we measured the patella height and average knee flexion angle during stance from the pre- and post-operative evaluations. Patella height and knee flexion angle were then used, along with the PTA simulation model, to predict the functional PTMA. The percent change in PTMA between pre-operative and post-operative valuations was computed.

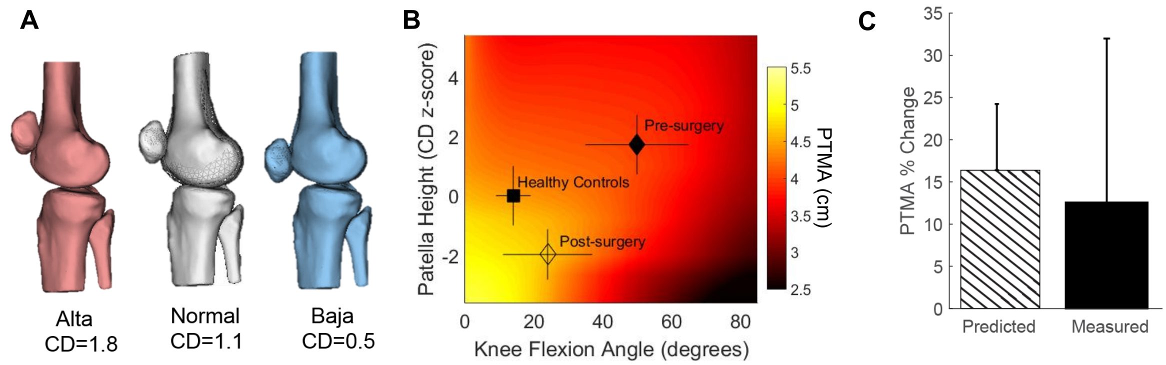

Figure 1: A) We modeled the distal translation of the patella tendon insertion as performed in a PTA procedure, then predicted the effects of patella height (*alta*, *normal*, and *baja*) on the patellar tendon moment arms, B) The model predicts that the PTMA is enhanced in extended postures when the patella moves from alta (CD z-score >1) to baja (CD z-score <1). Average and standard deviation of patella height (x-ray) and mean flexion angle in stance (gait) are shown for subjects that underwent the DFEO+PTA procedure, and from a healthy pediatric control group, C) A comparison of model-predicted and measured changes in the PTMA after combined DFEO/PTA surgery.

RESULTS

Our model of the PTA procedure suggests that the patellar tendon moment arm is highly dependent on patella position, with alta producing a greater PTMA in flexed postures and baja producing a greater PTMA in extended postures (Fig. 1B). The retrospective data show that PTA consistently induced a large change in patella height, moving it from substantial alta to substantial baja. The model predicted a 16% increase in PTMA as a result in the change in patella position, which was only slightly larger than the 13% increase in the PTMA measured in the patient population (Fig 1C).

DISCUSSION

The PTA procedure is used to tension the quadriceps, which is believed to modulate the quadriceps optimal operating lengths and thereby enhance quadriceps strength. Indeed, there is a strong tendency for subjects to exhibit greater knee extension and less quadriceps lag after undergoing PTA (and often DFEO) procedures [1]. However, our analysis suggests the improvement in knee extensor function may, in part, result from a larger patellar tendon moment arm that occurs from correcting patella alta. The model also suggests that baja could compromise the PTMA in more flexed postures. Further study is needed to see if this is true, and whether it might diminish an individual's ability to do functional tasks involving deeper knee flexion, such as chair rise.

REFERENCES

1. Novacheck, T et al. (2009) Journal of Bone and Joint Surgery, 91: 271:286.
2. O'Brien, T et al. (2009) Journal of anatomy, 215: 198-205.
3. Lenhart, R et al. (2015). Annals of Biomedical Engineering, 1-11.

ACKNOWLEDGMENTS

R21 HD084213

Comparison of Patellar Advancement Techniques in Children with Patella Alta and Cerebral Palsy

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Introduction

Crouch gait is defined as increased hip and knee flexion, and excessive ankle dorsiflexion throughout the stance period of gait.[1] Rarely observed in early ages, this movement abnormality affects children with cerebral palsy (CP) and progressively worsens as patients grow. Patellar advancement corrective surgery (PA) is used to treat the lack of dynamic knee extension seen in crouch gait in children with CP, however there is no consensus on the superiority of differing PA techniques for this population. This abstract compares surgical outcomes of proximal tendon resection (PT), patellar tendon imbrication (IM), and tibial tubercle advancement osteotomy (TT) with respect to gait and radiographic measurements in a cohort of individuals with CP.

Clinical Significance

Currently, there is sparse literature comparing the different PA surgical techniques that can be used to treat crouch gait, making this the first study comparing the different techniques. Our results show both radiographic and kinematic improvement from the PT procedure, which is most often used in our facility.

Methods

Subjects with a diagnosis of CP who underwent PA from 2000-2016 were identified. Adjusted Koshino Index (AKI), knee extension at mid-stance (KE_MSt), knee extension at terminal-swing (KE_TSw), and knee range of motion over the gait cycle (KROM) were collected from patients who had pre-op and post-op gait analyses and radiographs within 2-3 months of surgical intervention. AKI values reported are the number of standard deviations away from a normal Koshino Index at a given knee flexion. Group means comparison using paired t-tests was selected to compare preoperative and postoperative radiographic and gait changes for each type of PA procedure.

Results

There were 116 patients (58 Female, 78 Male) in which 139 extremities had a PA procedure. Of those, 107 received PT, 23 IM, and 9 TT. Gait data was available for 63 knees (50 PT, 10 IM and 3 TT). This facility defines a minimum change of 5 degree in KE_MSt, 5 degrees in KE_TSw, 10 degree in KROM, and 1.0 in AKI as clinically significant. All results are presented in Table 1.

Table 1

Measurement Outcome	Surgical technique	Pre-op, mean(SD)	Post-op, mean(SD)	Difference (post - pre), mean(SD)	p-value, paired t-test
Adjusted Koshino index of patellar height (n=139)	Proximal Tendon	3.92(4.46)	-0.14(4.59)	-4.06(4.51)	<0.0001*
Clinically Meaningful Difference AKI > 1.0	Imbrication	5.82(6.22)	0.65(2.73)	-5.17(5.53)	0.0002*
	Tibial Tubercle Advancement	3.61(4.57)	0.05(7.20)	-3.57(3.89)	0.0249*
Knee Extension at Mid-Stance (n=63)	Proximal Tendon	38.00(15.23)	22.74(14.57)	-15.26(15.87)	<0.0001*
Clinically Meaningful Difference KF_MSt > 5 degrees	Imbrication	31.05(17.41)	16.24(8.86)	-14.81(15.23)	0.0285*
	Tibial Tubercle Advancement	13.58(13.09)	4.71(19.16)	-8.88(8.85)	0.2243
Knee Extension at Terminal Swing (n=63)	Proximal Tendon	-9.69(10.23)	-4.33(9.34)	5.35(13.21)	0.0072*
Clinically Meaningful Difference KE_TSw >5 degrees	Imbrication	-10.00(7.84)	0.71(5.84)	10.71(9.58)	0.0011*
	Tibial Tubercle Advancement	0.00(8.66)	1.67(10.41)	1.67(2.89)	0.4226
Knee Range of Motion (n=63)	Proximal Tendon	27.99(10.56)	37.29(11.80)	9.30(10.59)	<0.0001*
Clinically Meaningful Difference KROM >10 degrees	Imbrication	27.14(12.47)	43.98(14.18)	16.84(9.25)	0.0013*
	Tibial Tubercle Advancement	43.64(18.05)	52.40(19.41)	8.77(10.17)	0.2742

Data Table of Outcome Variables: Statistically Significant($p<0.05$)=*

Discussion

All three surgical techniques showed evidence of improvement in all outcomes variables, although the TT technique did not reach significance due to small and unequal sample size. Typically showing successful results, PT is the primary technique used at our institution, explaining the limited number of IM and TT procedures available for this study. Upon comparison analysis between the three techniques, it was found that there was no statistical significant difference between the three techniques. Due to the limited number of IM and TT procedures, further analysis with expanded patient population, or a multicenter study, is necessary to compared and determine the optimal PA techniques for patients with CP.

References

1. Chang FM, Rhodes JT, Flynn KM, Carollo JJ. The role of gait analysis in treating gait abnormalities in cerebral palsy. *The Orthopedic clinics of North America.* 2010;41(4):489-506.

Disclosure Statement

The patient population, data, and analysis was all provided and completed at Children's Hospital Colorado.

CROUCH GAIT MODIFICATION DURING A SHORT WALKING EXERCISE

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INTRODUCTION

Crouch gait is characterized by excessive knee flexion and is one of the most common walking pattern in children with spastic diplegic cerebral palsy (CP). This gait induces muscle and knee joint stress [1], which increase with knee flexion. Hence, limiting excessive lower extremity flexion during walking may reduce secondary adaptations and preserve mobility function. Parent et al., recently reported a more severe crouch gait after a 6-minute walking exercise performed at comfortable speed [2]. These results suggested that during a short walking exercise, similar to daily life activities, children adopt a more severe crouch gait compared to the gait evaluated in clinical gait analysis (CGA). Early onset of muscle fatigue may be related to this gait modification. The present study aimed to specify when the kinematic adaptations occurred during a continuous walking exercise and to report the concomitant muscle activity.

CLINICAL SIGNIFICANCE

Evaluating gait progression during continuous walking exercise is relevant in order to consider in therapeutic decision making the gait adaptation that may occur in daily activities.

METHODS

To date, 9 children (2 females; mean age ± SD: 13 ± 2 years; body mass: 40.3 ± 8.4 kg; height: 155.9 ± 15.4 cm; GMFCS II-III) with bilateral spastic CP were included in the study. They had to have a knee flexion greater than 15° throughout the stance phase, and to be able to walk with or without walking aid for at least 6 minutes. Children were asked to walk barefoot (1) six times along a 10-meter walkway, as classically done in CGA, and (2) around a 25-meter pathway for 6 minutes at their self-selected speed. Standardized encouragements were provided at the third minute. Kinematic and electromyographic (EMG) data were acquired during CGA and at each minute of the 6-minute walking exercise (6mwe). Data from the second to the sixth minute were compared with the CGA data. The outcome measures included: (1) walking speed and step length; (2) hip, knee and ankle flexion (maximum, minimum, and range of motion (RoM)) during the single-limb stance; (3) center of mass (CoM) vertical position normalized with the height (maximum, minimum, and mean).

RESULTS

Walking speed ($p<0.05$) and step length ($p<0.05$) remained unchanged throughout the walking exercise except at the fourth minute (0.98 m/s+ 16% and 0.51m+ 10%), which corresponds to the measure following the standardized encouragements. The walking speed increased also significantly at the second minute (+18%, $p<0.001$) compared to the CGA speed. The maximal knee ($p<0.05$) and ankle flexion increased significantly at the sixth minute (see Fig. 1). The RoM of the knee increased significantly from the second to the sixth minute compared to the CGA ($p<0.05$). Compared to the CGA, minimal CoM vertical position decreased significantly from the second to sixth minute ($p<0.05$).

The mean CoM vertical position was significantly different from the second to the sixth ($p<0.05$), except at the fourth minute, which followed the standardized encouragements. To date, the EMG analysis did not show modification between CGA and the sixth minute of walking regarding the muscle co-activation.

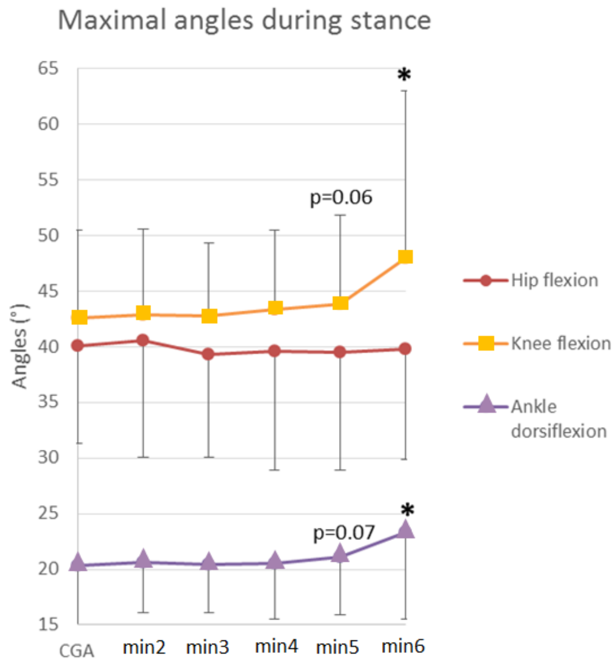

Fig 1: Joint angles during the walking exercise
*, significantly different from CGA, $p<0.05$

DISCUSSION

The children with CP who were included in the present study walked with a more severe crouch gait at the end of the walking exercise performed at self-selected speed. The more severe crouch gait appeared mostly at the end of the exercise. The ankle and the knee increased significantly during the sixth minute compared to GCA. However, the p-value of these two joints tend to be significant at the fifth minute ($p=0.065$ and 0.059, respectively). This study highlights that crouch gait increases after only few minutes of walking and, in turn, may lead to an increase in knee and ankle joints stress. The hip joint remained unchanged throughout the walking exercise. This result could be explained by trunk and pelvis compensations. The use of walk aid by 2 children have also probably increased the variability of the pelvis and the hip position within the group. The minimal vertical position of the CoM decreased significantly from the second to the sixth minute. As the children walk around the pathway, the minimal vertical position of the CoM decreased. The mean vertical position of the CoM, as well as the walking speed and the step length, were influenced by the standardized encouragements provided at the third minute. This would explain the significant differences found at these parameters at the fourth minute when compared to the CGA.

In conclusion, this study highlighted the importance to consider in therapeutic decision making the gait modifications that could occur after only few minutes of walking. The progression of crouch gait could be related to muscle fatigue. This parameter will be analyzed at each minute of the walking exercise based on the shift of the median frequency toward low frequencies.

REFERENCES

1. Steele KM (2012) Gait& Posture, 35(4): 556-60
2. Parent A (2016) Clinical Biomechanics, 34:18-21.

DISCLOSURE STATEMENT

The authors have no conflicts of interest to disclose.

RELATIONSHIP BETWEEN PATELLA HEIGHT, CROUCH SEVERITY, AND OUTCOMES FOLLOWING SURGICAL TREATMENT

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INTRODUCTION

Distal femoral extension osteotomy (DFEO) and patellar tendon advancement (PTA) surgeries are increasingly used to treat crouch gait in children. However, the surgical approach and indications for the procedures continue to evolve [1], with a shift towards using the treatment on younger patients. A primary question that remains is how much to advance the patellar tendon? After a DFEO, the primary goal of PTA is to re-tension the quadriceps and address quadriceps insufficiency. Yet, the current target for PTA is anatomical, and typically involves advancing the patella from alta into baja. In this study, we first investigated the relationship between patella alta and knee flexion in stance, to gain a better understanding of how PTA magnitude may scale with crouch severity. We then retrospectively investigated whether outcomes were related to age and post-operative changes in knee geometry and behavior.

CLINICAL SIGNIFICANCE

The results of this study support the use of age when considering DFEO and PTA procedures to treat crouch. Individuals with good outcomes were significantly younger, exhibited larger reductions in knee flexion contractures and displayed greater enhancement of the patellar tendon moment arm after surgery.

METHODS

Subjects: We used the clinical gait database at Gillette Children's Specialty Healthcare)St. Paul, MN) to identify patients with a diagnosis of cerebral palsy who underwent DFEO + PTA procedures after 2002. A total of 39 patients (29 male, age=13±3 years) with 63 knee joints (33 left, 30 right) fulfilled our criteria. For each subject, we collected the minimum knee flexion in stance (crouch severity) from pre- and post-operative gait analyses. Good outcomes ("Improved" in Table 1) were classified as a post-operative change in knee flexion angle during stance greater than 15˚ or minimum knee flexion angle during stance that was within the 95% confidence interval of normative pediatric values (<18˚).

***Knee Geometry and Function*:** Patella height was measured using the Caton-Deschamps (CD) index and is reported as a z-score relative to healthy adults. Patellar tendon moment arms (PTMA) were calculated from lateral x-rays by finding the perpendicular distance from the patellar tendon to the geometric center of the femoral condyles [2]. PTMA values were scaled to the epicondylar width. The tibial slope was the angle between the anterior cortex of the tibial shaft and the posterior aspect of the transverse line approximating the epiphysis of the proximal tibia [3]. Knee flexion contracture and extensor lag were measured as part of the standard physical exams at pre- and post-operative gait collections.

Statistical Analysis: Univariate correlations were performed between minimum stance phase knee flexion and CD index. Student t-tests (with Bonferroni post-hoc corrections) were used to test for significant changes in knee geometry and function metrics between the pre- and post-operative evaluations. Effect of age, knee geometry and knee function on surgical outcomes were assessed using a mixed effects logistics regression. Significance was set at $p<0.05$.

RESULTS

There was a significant correlation (R^2: 0.2 conditional, 0.6 marginal) between patella height and knee flexion in the pre-operative gait analysis, but this relationship was no longer evident post-operatively (Fig. 1). After surgery, there was a significant decrease in knee flexion contracture, extensor lag, patella height, and tibial slope and an increase in the PTMA. There were 54 and 9 subjects who met our criteria for good ("Improved") and poor ("Unimproved") outcomes, respectively. Subjects who exhibited good outcomes were significantly younger, had a greater reduction in knee flexion contracture and had significantly greater enhancement of the patellar moment arm after surgery (Table 1).

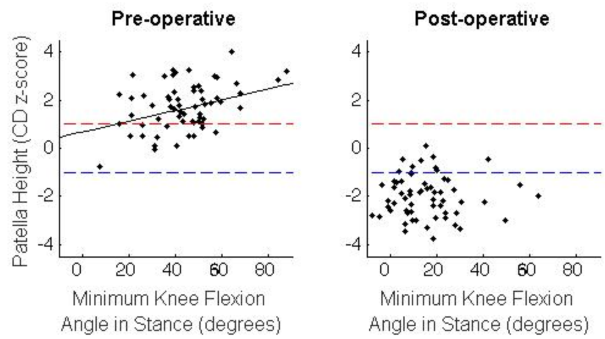

Figure 1: Patella height, measured by the CD Index, was correlated to crouch severity pre-operatively but not post-operatively. Red and blue dashed lines designate alta and baja patella positions, respectively.

Table 1: Mean (±1 SD) changes in select knee geometry and function metric between pre- and post-operative evaluations. Improvements in gait were related to age and post-surgical changes in contractures and PTMA.

Metric	All	*p*	Improved	Unimproved	*p*
Age at surgery	--	--	14±3	18±4	0.002
Δ Knee Flexion Contracture (degrees)	-6±3	0.0001	-16±12	-5.6±11	<0.0001
Δ Extensor Lag (degrees)	-9±4	<0.0001	-7.7±14	-14±17	0.8
Δ Patella Height (CD z-score)	-3.7±0.2	<0.0001	-3.8±1	-3.4±0.9	0.7
Δ PTMA (% change)	12±4	<0.0001	16±18	-8±14	0.01
Δ Tibial Slope (degrees)	-2±1	0.008	-1.4±4	-5.1±5	0.2

DISCUSSION

PTA is standardly performed in a way that aligns the inferior pole of the patella with the knee joint line [4]. This anatomical target means that the patellar tendon advancement is larger for those subjects with greater patella alta. This study shows that patella height and knee flexion in gait are weakly correlated prior to treatment, such that advancement magnitude generally scales with crouch severity. However, the lack of correlation between patella height and crouch after surgery suggests that baja positions may not be necessary to achieve good outcomes. Our results show that age was a significant delineator of good and poor outcomes, which supports the trend toward the use of the treatment on younger patients. Good outcomes were also associated with larger reductions in knee flexion contractures and greater enhancement of the PTMA, such that it is possible that skeletal growth after surgery may have been a factor in the outcomes.

REFERENCES

1. Stout, J et al. (2008) J Bone Joint Surg Am, 90(11): 2470-2484
2. O'Brien T et al. (2009) Journal of Anatomy, 215: 198-205
3. Patthanacharoenphon C et al. (2013) Journal of Children's Orthopaedics, 7: 139-146.
4. Novacheck, T et al. (2009) Journal of Bone and Joint Surgery, 91: 271:286.

ACKNOWLEDGEMENTS

We thank Scott Hetzel for help with statistical methods. Work is funded by NIH R21 HD084213.

DOES DISTAL FEMORAL EXTENSION OSTEOTOMY WITH PATELLAR TENDON ADVANCEMENT IN INDIVIDUALS WITH CEREBRAL PALSY HELP IN THE LONG-TERM?

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INTRODUCTION

Short-term gait improvements are achieved after a distal femoral extension osteotomy with patellar tendon advancement (DFEO+PTA) in individuals with cerebral palsy (CP) who walk in crouch [1], but long-term outcomes are unknown. We measured long-term outcomes (body structure/function, activity, participation, and pain) in individuals who had a DFEO+PTA (cases) compared to baseline and individuals who did not have a DFEO+PTA (controls). Secondly, we ascertained if short-term gait improvements are maintained at long-term.

CLINICAL SIGNIFICANCE

Knowing the comprehensive long-term outcomes after DFEO+PTA will allow for more complete counseling of patients who walk in crouch and aid in post-surgical goal setting.

METHODS

This was an IRB approved cohort study. All participants had CP, walked in crouch (knee flexion angle >2 SDs above typically developing at initial contact and minimum flexion) and had knee flexion contracture(s) ≥10° at baseline, were ≥20 years old at long-term, and ≥8 years post-DFEO+PTA (cases). Controls had no or alternative treatments to DFEO+PTA. At long-term, all participants completed questionnaires (quality of life, satisfaction with life, functional assessment questionnaire, frequency of participation, functional mobility scale, pain) and a subset completed a gait analysis, 5-times sit-to-stand (5xSTS), and timed-up-and-go (TUG). The primary outcomes were knee flexion/extension at initial contact and minimum flexion. Wilcoxon rank sum, signed rank, and chi-square tests were used (α=0.05).

RESULTS

Of the 66 participants, 52 returned for an analysis (remainder only completed questionnaires). An additional short-term gait analysis was available for 13/28 cases. At baseline, cases had more abnormal gait and higher oxygen consumption than controls ($p<0.05$, Table 1). Between baseline and long-term, knee flexion angles did not change for controls, whereas knee flexion angles improved for cases (Table 1). From short- to long-term, cases lost 6.1° (p=0.004) and 5.9° (p=0.007) of knee extension at initial contact and minimum flexion, respectively. Among case and control limbs, 36% and 61%, respectively, were in crouch at long-term (p=0.045). The 5xSTS was completed by 42% of each group. Controls tended to perform it faster (p=0.18). There were no differences in TUG or questionnaire responses.

DISCUSSION

A DFEO+PTA improves knee extension during gait into adulthood for most individuals, which is superior over no or alternative treatment(s). On average, however, the superior knee kinematics did not affect life satisfaction, activity, participation, or pain relative to controls.

Table 1. Comparison between case and control groups that completed baseline and long-term follow-up analyses (median(IQR)).

	Ambula-tory, n (%)	**Assistive device**, n (%)	**Limbs** (n)	**Age at gait analysis** (yrs)	**FAQ**	**GDI**	**Knee flexion θ IC** (°)	**Min knee flexion θ** (°)	**Peak knee extensor moment** (ND)	**Knee flexion contracture** (°)	**O_2 cons** (ND % speed-matched TD)	**TUG** (sec)	**5x STS** (sec)
BASELINE													
Case	28/28 (100%)	15/28 (54%)	40	13.5 (4.3)	7(3)	59(9)[B]	42.4 (13.7)[B]	38.3 (10.7)[B]	0.100 (0.040)[B]	15(10)[B]	375 (147)[B]		
Control	24/24 (100%)	11/24 (46%)	34	13.1 (2.3)	8(2)	68(11)[B]	36.1 (9.6)[B]	27.1 (10.3)[B]	0.083 (0.034)[B]	10(5)[B]	263 (172)[B]		
Effect size (γ_1^*) between groups					0.00	0.84	-0.52	-1.64	-1.44	-1.97	-0.82		
LONG-TERM													
Case	27/28 (96%)	15/27 (56%)	40	25.9 (6.3)	8(4)	64(14)[C]	28.4 (13.3)[C]	11.9 (18.8)[L,C]	0.069 (0.037)[C]	0(5)[L,C]	253 (64)[C]	17.1 (13.7)	20.3 (10.1)
Control	22/24 (92%)	11/22 (50%)	34	27.4 (7.0)	8(4)	62(13)[C]	35.2 (19.5)	21.1 (20.2)[L]	0.066 (0.044)	10(8)[L,C]	254 (139)	15.0 (20.9)	14.4 (8.9)
Effect size (γ_1^*) between groups					-0.09	-0.10	0.92	0.58	-0.74	1.04	-0.13	-0.14	-0.97
TD reference (mean±SD)					10	100±10	3.9±5.7	6.0±6.1	0.043± 0.013	0	100		

cons: consumption; **FAQ**: functional assessment questionnaire; **GDI**: gait deviation index; **IC**: initial contact, **ND**: non-dimensional; **TD**: typically developing; **TUG**: timed up-and-go; **5x STS**: 5-times sit-to-stand. [B]significant difference between groups at baseline ($p<0.05$), [L]significant difference between groups at long-term follow-up ($p<0.05$), [C]significant difference from baseline values ($p<0.05$)

REFERENCES

1. Stout, J. et al. (2008) J Bone Joint Surg Am, 90(11):2470-2484.

DISCLOSURE STATEMENT No author has any conflicts of interest to disclose.

SURGICAL TREATMENT OF PES PLANOVALGUS IN AMBULATORY CHILDREN WITH CEREBRAL PALSY: EFFECT ON GAIT AS CHARACTERIZED BY MULTI-SEGMENT FOOT MOTION ANALYSIS AND FOOT DEFORMITY

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INTRODUCTION

Foot deformity and pathologic gait are frequently observed in ambulatory children with cerebral palsy (CP). Weakness, poor motor control and abnormal tone associated with CP leads to altered loading and structural integrity of the foot. Most commonly, pes planovalgus develops and is associated with changes in gait. Surgical intervention is indicated when bracing is ineffective with the main goal of creating a stable plantigrade foot for standing and walking.

Pes planovalgus is characterized by forefoot varus and a hypermobile collapsed midfoot. Forefoot varus is accompanied by hindfoot eversion and midfoot abduction during gait while the collapsed midfoot results in midfoot dorsiflexion. Physical examination and radiographic measures are utilized in defining the pes planovalgus deformity, while the multi-segment foot model is an emerging standard for characterizing gait with respect to hindfoot and midfoot motion [1].

This study examines the effect of flatfoot reconstruction on (1) physical examination and radiographic measures of deformity and (2) gait utilizing multi-segment foot motion analysis. We hypothesize that flatfoot reconstruction improves forefoot varus and midfoot collapse associated with pes planovalgus as quantified by physical examination findings in the subtalar joint neutral (STJN) position, radiographic parameters, and multi-segment foot motion analysis.

CLINICAL SIGNIFICANCE

This study defines the outcome of flatfoot reconstruction in ambulatory children with CP with regard to foot deformity and gait.

METHODS

After institutional review board approval, ambulatory children with CP undergoing flatfoot reconstruction at our institution were identified. All surgical patients underwent calcaneal lengthening with either first ray dorsal opening wedge osteotomy or plantar closing wedge osteotomy. Surgery was performed by one of nine fellowship trained pediatric orthopaedic surgeons. Patients included in the study underwent preoperative and postoperative multi-segment foot motion analysis in our institution's gait lab based on a model developed by Leardini et al [1]. In addition, non-surgical ambulatory patients with CP and pes planovalgus were identified as controls. All patients in the control group underwent gait analysis at two separate time points.

The following motion analysis variables were compared preoperatively and postoperatively in the surgical group and at two separate time points in control group: mean hindfoot eversion with respect to the tibia, maximum midfoot dorsiflexion, and maximum midfoot abduction, all during stance. In addition, physical examination findings (weightbearing and nonweightbearing STJN position of the forefoot, midfoot and hindfoot) and radiographic parameters on standing radiographs (AP and lateral talar – first metatarsal angles and calcaneal pitch) were utilized to characterize changes preoperatively and postoperatively and at the two gait analysis sessions in the control group. Frequencies for categorical variables and means and standard deviation for continuous variables were calculated. One-sided t-tests were performed to compare pre-operative and post-operative values for continuous variables.

RESULTS

24 surgical and 17 control patients were identified (Table 1). In total 42 flatfoot reconstructions were performed with 18 subjects undergoing bilateral procedures. We first assessed forefoot varus. 83.3% of feet were characterized as having forefoot varus in the STJN position preoperatively compared to 45.2% postoperatively. Decreased STJN offset of the midfoot in the coronal plane ($p<0.001$) as well as decreased mean AP talar – first metatarsal angle on standing radiographs ($p<0.001$) were also observed. These findings were associated with the following changes in gait: decreased mean hindfoot eversion ($p=0.005$) and decreased maximum midfoot abduction ($p=0.002$). No changes were noted in these variables for the control group.

Changes in the midfoot were also found with surgery. On physical examination, 90.5% of feet were characterized as planus with weightbearing preoperatively compared to 50.0% postoperatively. In addition, calcaneal pitch increased ($p=0.002$) and the lateral talar – first metatarsal angle decreased ($p<0.001$) postoperatively on standing radiographs. On gait analysis, midfoot dorsiflexion decreased postoperatively ($p<0.001$). Again, no changes were noted in these variables for the control group.

Table 1: Demographics for surgical and control groups.

	Surgical group (n=24)	**Control group (n=17)**
Male	14 (58.3%)	11 (64.7%)
Age at first gait analysis	9.7 ± 3.4 years*	10.3 ± 3.5 years
Age at second gait analysis	11.2 ± 2.3 years**	11.2 ± 3.6 years

*Preoperative gait analysis; **Postoperative gait analysis

DISCUSSION

We now know that flatfoot reconstruction in ambulatory children with CP improves forefoot varus and midfoot collapse as measured by physical examination and radiographic parameters. Correction of forefoot varus and midfoot collapse with surgery is manifested in improvements in gait parameters observed in multi-segment foot motion analysis, including hindfoot eversion and midfoot abduction and dorsiflexion during stance. These changes were not observed in the control group, suggesting that improvement in forefoot varus and midfoot collapse are not simply due to the natural history of pes planovalgus. These data support the efficacy of flatfoot reconstruction in ambulatory children with CP. Future study will assess the validity of the multi-segment foot model to physical examination and radiographic parameters.

REFERENCES

1. Leardini, A. et al. (2007) Gait Posture, 25(3):453-62.

DISCLOSURE STATEMENT

The authors have no disclosures.

THE PERIACETABULAR OSTEOTOMY IMPROVES RADIOGRAPHIC AND GAIT FUNCTIONAL OUTCOMES OF ADOLESCENTS WITH CEREBRAL PALSY

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INTRODUCTION

Reports on the ability of the Ganz Periacetabular Osteotomy (PAO) to correct severe hip dysplasia in patients with underlying neuromuscular syndromes are limited. Previous studies have primarily focused on radiographic outcomes alone. The purpose of this prospective case series was to assess the radiographic, self-reported function and gait outcomes in patients with cerebral palsy (CP) who present for surgical correction of severe hip dysplasia

CLINICAL SIGNIFICANCE

The Ganz PAO has been previously shown to radiographically correct severe hip dysplasia however there are no reports of functional outcomes when performed in adolescent ambulatory patients with cerebral palsy.

METHODS

This is an IRB-approved prospective analysis of patients with CP treated with a PAO for severe hip dysplasia. Standard radiographic measures included the lateral center edge angle (LCEA), acetabular index (AI) and ventral center edge angle (VCEA). Self-reported functional was assessed using the modified Harris Hip Score (mHHS, max score 89.) Lower extremity kinematics were assessed using a modified Helen Hayes marker set and processed using Vicon Nexus (Plug-in-gait). The gait deviation index (GDI) was determined using a representative walking trial of the affected side(s). Pre and post-op variables were compared using a Student t-test ($p<0.05$).

RESULTS

Nineteen patients (18 Males/1 Female) with 20 subluxated hips (9 Right/11 Left) were included in the analysis at an average age at the time of surgery of 16.5±2.1 yrs (range, 13.3-20.4 yrs). Concomitant procedures included soft tissues procedures (psoas lengthening and/or adductor tenotomy/release, n=11), proximal femoral valgus (n=8) and derotational (n=1) osteotomies. Average length of follow-up for all 19 patients was 2.2±0.9 yrs (range, 0.9-4.7yrs). The group included 10 hemiplegic, 6 diplegic, 1 triplegic, and 2 quadriplegic patients.

There were seven patients for whom pre-op data was not included in the analysis: two were non-ambulatory, three completed gait analysis greater than 6 mos prior to surgery (range 1 to 4 years previous to surgery) and two patients were not tested. Two data analyses were conducted: a) pre and post-operative comparison of the 13 subjects who were tested pre and post-operatively, and b) post-operative description only of all 20 patients.

Post-operatively, maintenance of walking was achieved in all subjects, and the 2 acute non-ambulators resumed walking. Radiographic parameters improved from pre-op to final follow-up: LCEA (-9° to 25°, $p<0.001$), AI (26° to 10°, $p<0.001$), VCEA (-12° to 27°, $p<0.001$). The GDI for the 13 patients who completed gait analysis pre-operatively was 46.7±13.4 (range, 27.2-68.7) pre-operatively and 48.4±17.0 post-operatively (range, 16.1-79.5),p=0.711). Within this cohort, 3/13 patients (23%) increased GDI post-operatively (average improvement: 18 points), 8/13 patients had no clinical change in GDI, and two patients decreased GDI (average decline: 8 points). At final follow-up, the average GDI for all patients was 46.1±16.5 (range, 16.1-79.5). The average walking speed was 1.0±0.3 pre-operatively and 0.9 ±0.2 post-operatively (p=0.774). The average post-operative mHHS for all subjects was 74.6 ±14.6 (28.0 to 89.0). For those with pre and postop surveys, the mHHS significantly improved 10 points, from 69.1 to 79.4 (p=0.020).

Table 1: Radiographic outcomes (N=20 hips)

	LCEA	AI	VCEA
Pre-op:	-9 ± 16°	26 ± 8°	12 ± 14°
Post-op:	25 ± 10°	10 ± 7°	27 ± 10°

Table 2: Self-reported function and gait outcomes for a) those with pre- and post- operative gait assessments, b) all patients with post-operative assessment

	mHHS	Walking Speed	GDI
a) N=13 hips (have pre and post assessment)			
Pre-op	69.1 ± 11.9	1.0 ±0.3	46.7±13.4
Post-op	79.4 ± 7.6 **	0.9 ±0.2	48.4±17.0
b) N=20 hips (all patients post assessment) †			
Post-op	74.6 ±14.6	0.9 ±0.2	46.1±16.5

* significantly improved post-operatively ($p<0.05$)
† includes 7 patients who were not tested pre-operatively

DISCUSSION

The Ganz PAO for severe hip dysplasia in patients with CP significantly improves radiographic parameters and results in maintenance of gait parameters at follow-up and may allow those acutely non-operative patients to become ambulatory. The goal of this Ganz PAO in this patient population is to provide a radiographically and clinically stable hip joint with a reduction in pain. Functional gains seen may demonstrate continued improvements with further follow-up in patients with neuromuscular conditions. The Ganz PAO is effective in treating severe hip dysplasia/subluxation in adolescent patients with CP and improves self- reported outcome scores.

REFERENCES

1. Ganz et al. *Clin Orthop*1988;232:26-36

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DISCLOSURE STATEMENT

The authors have no conflicts of interest to disclose.

Rate of Force Development in Isometric Strength Tests are Related to Self-reported Physical Activity in Adults with Cerebral Palsy

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INTRODUCTION

Successful performance of most functional tasks relies on coordinated movements that in turn require skeletal muscles to generate sufficient strength and speed. While muscle weakness is not uncommon for individuals with cerebral palsy (CP), even those with sufficient strength may have difficulty accomplishing tasks if they cannot achieve necessary force levels quickly. This suggests the *rate of force development* (RFD) may be equally critical to daily activities as strength. Previous investigators have reported on this phenomenon in children with CP [1]. This study aims to determine if this trend is also observable in adults with CP, using self reported outcome measures of physical activity from a validated instrument.

CLINICAL SIGNIFICANCE

Understanding the relationship between rate of force development and physical activity in adults with CP may provide a useful biomarker of functional ability suitable for use in the clinic or motion laboratory. It may also help identify interventions that can address limitations in this area, leading to improved performance in tasks that require both strength and speed.

METHODS

This project was part of a larger federally funded prospective study (Cerebral Palsy Adult Transition Study - CPAT) designed to understand the relationship between walking ability, overall health status, and risk for secondary health conditions. A preliminary cohort of 36 young adults with CP (GMFCS level I-III, mean age = 24.1 years, mean GDI = 76.1, affected/non-dominant side) who previously had Instrumented Gait Analysis (IGA) at our facility as children were included. Isometric knee extension strength at 60^o of knee flexion (85^o seated hip flexion) using a HUMAC NORM isokinetic dynamometer were tested bilaterally using 3 maximum contractions from rest, sustained for 5 seconds. Maximum average RFD (MA-RFD) were calculated from the slope of moment/time curves at intervals from 0 to 30, 50, 100, and 200 ms, with Peak RFD values determined from the highest average slope attained from the 3 trials. Correlation analysis and paired t-tests were used to compare RFD variables to selected physical function outcomes scores from the PROMIS-57 physical function domain.

RESULTS

A significant positive correlation between Peak RFD of the knee extensors and the physical function total score of the PROMIS-57 were identified using both the Spearman and Pearson correlation coefficients in this initial cohort as shown in the table below and Figure 1.

DV	IV	Spearman r	p for Spearman r	Pearson r	p for Pearson r	p<0.05
Peak RFD	PROMIS 57: physical function total score	0.41	0.016	0.40	0.017	Y

While results from both lower limbs were tested, a significant difference was only seen on the more-affected / non-dominant side.

DISCUSSION

This study found a significant positive correlation between a validated measure of self-reported physical function, and the rate of force development (RFD) of the quadriceps muscle using a simple, isometric, knee extension task in adults with CP. Ambulatory subjects able to more rapidly generate force during isometric contraction of the quadriceps on their weaker side reported higher levels of physical functioning on the PROMIS-57.

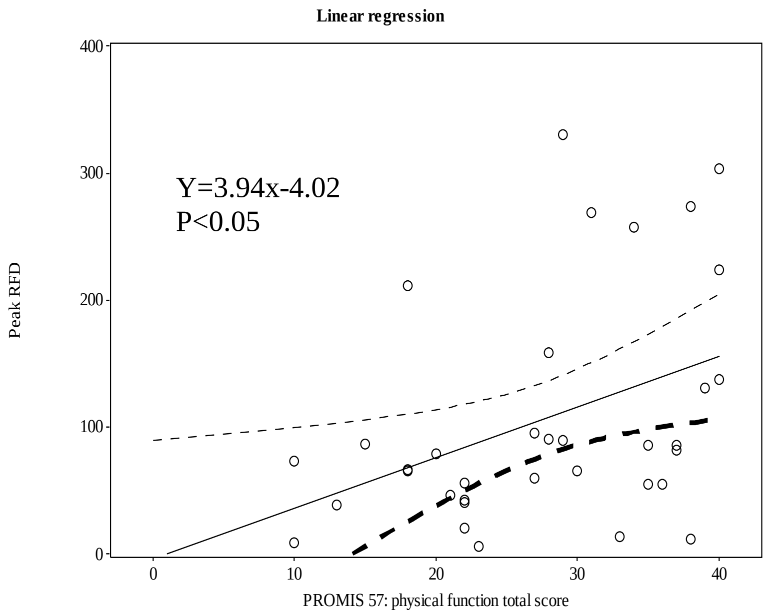

Figure 1: Comparison between PROMIS57 and Peak

These results from an adult cohort are consistent with the results reported in a previous study that examined peak RFD in children with CP [1]. This cohort's peak RFD was lower than that measured in the normal population from the previous study, but higher than reported for children with CP. The finding that only RFD data from the weaker, less affected, or non-dominant side were significant, despite the cohort's inclusion of individuals with both hemiplegic and diplegic CP, suggests limitations on the weaker side have the greatest influence on overall physical function.

CONCLUSION

We believe this preliminary study supports the notion that RFD can be used as a quantitative, physical measure of functional ability, and suggest that both muscle strength and speed are important biomarkers of physical function in adults with CP. We are encouraged by these results, as the simplicity of isometric testing does not necessitate the use of an isokinetic dynamometer, and likely can be reproduced with less specialized quantitative devices in the clinic.

ACKNOWLEDGEMENTS

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References

1) Moreau NG, Falvo M, Damiano DL. Rapid Force Generation is Impaired in Cerebral Palsy and is Related to Decreased Muscle Size and Functional Mobility. *Gait & posture*. 2012;35(1):154-158. doi:10.1016/j.gaitpost.2011.08.027.

DISCLOSURE STATEMENT: The authors have no conflicts of interest to disclose.

Poster Session #1: Odd Posters

	TITLE	AUTHORS
1	Movement Differences Between Children with Autism and Children with Typical Development: Evidence for Evaluating the Individual Before the Group	Janet Dufek, Jeffrey Eggleston, John Harry
3	Reliability Of The Footwear Total Mediolateral Asymmetry Score Tool	Codi Ramsey, Daniel Cury Ribeiro, Peter Lamb, Christopher Charles Sole, Gisela Sole
5	A Comparison of In Vivo Tibiotalar and Subtalar Kinematics in Chronic Ankle Instability Patients and Asymptomatic Controls Using High-speed Dual Fluoroscopy	Koren Roach, K. Bo Foreman, Alexej Barg, Andrew Anderson
7	Quantitative Assessment of Walker Use During Gait	Sean Brown, Jon Davids, Anita Bagley
9	Gait and Mobility in a Multigenerational Southeast Asian Family With Type I Osteogenesis Imperfecta	Nikhil Kurapati, Thaddeus Rogozinski, Rebecca Boerigter, Carlo Sumpaico, Joycie Abiera, Melanie Alcausin, Peter Adamczyk, Peter A. Smith, Gerald F. Harris
11	Focal Dystonia: A Case Study	Rachel Binkley-Vance, Roy Davis, Tamara Fatianov; Michael Mendelow, David Westberry,
13	Improved Hip Joint Center Prediction	Emily Miller, Kenton Kaufman
15	Soft Tissue Artifact Leads to Errors in Biomechanical Model Outputs at the Hip Joint	Niccolo Fiorentino, Penny Atkins, Michael Kutschke, K. Bo Foreman, Andrew Anderson
17	Importance of Subject Specific Kinematics and Kinetics for Predicting Cartilage Contact Mechanics	Penny Atkins, Niccolo Fiorentino, Samuel Colby, Andrew Anderson
19	Effects of Cognitive Tasks and Depth Perception on Elderly Balance on a Stable and Two Passively Unstable Surfaces	Peter Quesada, Coty Groeschen, Benjamin Mueller
21	A Three-dimensional Analysis Of Ilium Innominate Movements to Evaluate a Clinical Diagnostic Test for Sacroiliac Joint Pain.	Beth Moody Jones, Michael Kurita, Natalie Fan, Yuri Yoshida
23	Withdrawn	
25	Therapies and Bracing for the Treatment of Crouch Gait in CP: A Case Study	Sylvia Ounpuu, Jennifer Rodriguez-MacClintic, Kristen Pierz
27	Sub-threshold Vibration for the Enhancement of Sensation and Function In Transtibial Amputees: Preliminary Results	Jenny Kent, Kota Takahashi, Zachary Meade, Abderrahman Ouattas, Nicholas Stergiou
29	Effect of Unilateral Hip Orthosis Stiffness on Gait in Healthy Individuals	Yusuke Sekiguchi, Dai Owaki, Keita Honda, Kenichiro Fukushi, Noriyoshi Hiroi, Takeo Nozaki, Shin-Ichi Izumi
31	Walker Assisted Amputee Gait Analysis Using a Mobile Prosthesis Integrated Sensor System	Omid Jahanian, Alyssa Schnorenberg, Joel Kempfer, Barbara Silver-Thorn, Brooke Slavens

Poster Session #1 (continued)

	TITLE	AUTHORS
33	Effects of Student-led Balance Training on a Community-Dwelling Older Adult with Recurrent Falls: A Case Study	Tzurei Chen, Janae Wagler
35	The Feasibility and Efficacy of a Standing Serial Reaction Time Task to Measure Postural Motor Learning	Genevieve Olivier, Christopher Walter, Kirsten Gorski, Sydney Schaefer, Serene Paul, Lee Dibble
37	Effects of a Novel Two-phase Rehabilitation Program on Postural Control in Older Adults	Evan Papa, Mahdi Hassan, Sandra Hunter, Rita Patterson, Nicoleta Bugnariu
39	Gait Outcomes of Children with Clubfeet Treated with Botulinum Toxin-A	Alec Black, Jessica Maurer, Karen Davies, Richard Beauchamp, Christine Alvarez
41	Impact of Glenohumeral Joint Congruity on Shoulder Function in Brachial Plexus Birth Palsy	Tyler Richardson, Elizabeth Rapp, Ross Chafetz, Stephanie Russo, Dan Zlotolow, Scott Kozin, James Richards

MOVEMENT DIFFERENCES BETWEEN CHILDREN WITH AUTISM AND CHILDREN WITH TYPICAL DEVELOPMENT: EVIDENCE FOR EVALUATING THE INDIVIDUAL BEFORE THE GROUP

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INTRODUCTION

Autism spectrum disorder (ASD) is a network of complex disorders affecting one in 68 children [1], resulting in healthcare costs that exceeds $3 million over the lifespan [2]. Although ASD is generally characterized by social and communicative deficits, contemporary evidence indicates ASD is also characterized by motor impairments, particularly walking [3]. A primary concern in this population is that such impairments can interfere with a child's motivation to engage in social activities [3,4]. Motor impairments in this population have yet to be consistently identified in the literature, and the impairments identified have proven difficult to generalize to the greater population of children with ASD. It is likely that the inconsistent empirical observations and the challenge to generalize impairments are due to the heterogeneous physical manifestations of the disorder [5].

The failure to generalize walking impairments in children with ASD appears due to the assumption of homogeneity in the traditional statistical approaches that dominate the literature. Although it can be assumed that data from a sample of children with TD will be homogeneous, the same is not true for children with ASD due to the distinct characteristics exhibited by each child with ASD [5]. Performing a group analysis on children with ASD can conceal the individuality of each child, producing an average value for the group that represents no single participant in the sample [6]. Therefore, the purpose of this study was to evaluate walking mechanics between children with ASD and children with TD using outcomes from a traditional group statistical approach and outcomes from a novel approach designed to account for participant heterogeneity.

CLINICAL SIGNIFICANCE

Traditional statistical approaches may not provide reliable representations of a clinical population when compared to populations with TD, illustrated here using children with ASD. Individual-specific statistical approaches can provide researchers and clinicians with a more thorough understanding of motor dysfunction within a clinical population. In turn, more appropriate interventions can be implemented to address the distinct features of each individual.

METHODS

Ten children with a clinical diagnosis of ASD (14 males, 6 females; 9.0±2.2 yrs 1.4±0.2 m, and 34.8±13.4 kg) and ten children with TD (9.0±2.1 yrs, 1.4±0.1 m, and 35.7±10.2 kg) participated in this study. Each child with TD was age- and gender-matched to a child with ASD. Each participant performed 20 over-ground walking trials at a self-selected speed (1.27± 0.22 m/s, ASD; 1.30 ± 0.18 m/s, TD). Three-dimensional kinematic data were obtained bilaterally using an 8-camera motion capture system (120 Hz). Data were smoothed using a low-pass Butterworth digital (6 Hz cutoff). Joint angular positions of the hip, knee, and ankle joints were calculated, and ensemble mean and standard deviation-time histories were calculated for each joint. All data were normalized to 100% of the gait cycle (101 data points).

To compare the ASD group to the TD group using a traditional group statistical approach, mean values were calculated for each participant for each joint. Mean values were then calculated for the ASD and TD groups, respectively. Independent t-tests ($\alpha = 0.05$) were used to compare the group means at each of the 101 data points of the gait cycle. The Model Statistic [6] technique ($\alpha = 0.05$) was utilized, in a similar fashion, to compare individual children with ASD to the TD group. For each approach, an equal number of trials/group observations (n=10) were used for the individual/group comparisons.

RESULTS

Few significant differences were revealed using the group analysis (left panel of Figure 1). However, a substantial number of significant differences were observed when individual children with ASD were compared to the TD group (middle and right panels of Figure 1).

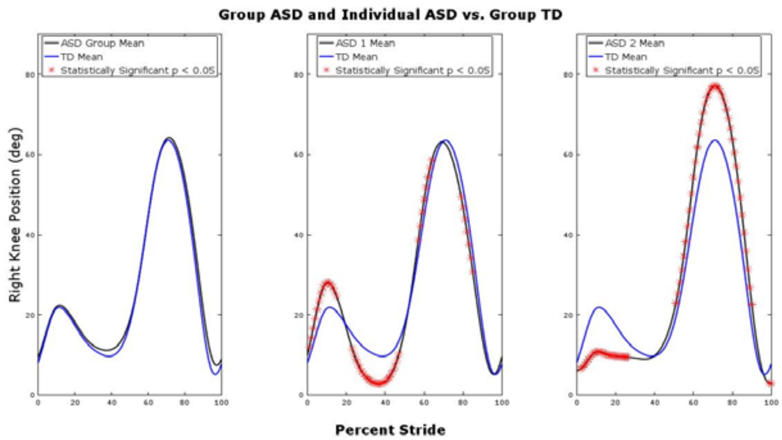

Figure 1. Comparison of group ASD vs TD, and individual children with ASD vs the TD group.

DISCUSSION

The current data indicate that it may be more appropriate to examine movement-abilities in children with ASD on an individual basis. The group statistical design masked the unique movement differences exhibited by each child with ASD, thereby diminishing the importance of each individual [6]. In conclusion, researchers and clinicians should proceed with caution when selecting a statistical design for evaluating movement abilities in children with ASD. The statistical approach selected should evaluate the movement characteristics of the individual before generalizing the results of the individuals to the greater ASD population.

REFERENCES

1. Autism Spectrum Disorders (2016). http://www.cdc.gov/ncbddd/autism/data.html.
2. Zablotsky B, et al. (2015). Natl Health Stat Report 2015, 87:1-20.
3. Kindregan D, et al. (2015). Autism Research and Treatment 2015.
4. Campione G, et al. (2016). Journal of Autism and Developmental Disorders, 46(6):1985-99
5. Brincker M & Torres E, (2013). Frontiers in Integrative Neuroscience, 7,34.
6. Bates B et al. (1996). In N. Stergiou (Ed.), Innovative Analyses of Human Movement, 3-28.

ACKNOWLEDGMENTS

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DISCLOSURE STATEMENT

The authors have no conflicts of interest to disclose.

RELIABILITY OF THE FOOTWEAR TOTAL MEDIOLATERAL ASYMMETRY SCORE TOOL

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INTRODUCTION

Footwear design and degradation contribute to asymmetric patterns between the medial and lateral aspects of the soles [1]. For example, motion control shoes are designed with an increased density at the medial midsole and materials such as ethylene vinyl acetate (EVA) degrade and deform with repetitive loading and friction with the foot-ground interface. Previous tools used to evaluate footwear components have not considered the role of this mediolateral asymmetry. The footwear total mediolateral asymmetry score tool (TAS) [1] to measure the height and density differences between the medial and lateral aspects of the shoe sole. TAS identifies the location and magnitude of height and/or density discrepancies of the inner-, mid- and outer sole. Only one study has assessed the mediolateral asymmetry [2], which found that healthy controls had larger asymmetries at the lateral rearfoot and higher plantar pressure profiles compared to participants with a history of lateral ankle instability. However, the reliability of the TAS has not been established, therefore, the aim of this study was to determine the within-rater, between-rater and between-day reliability of the footwear TAS tool.

CLINICAL SIGNIFICANCE

The TAS provides an objective measurement of the condition of the patient's shoes. Future research is needed to determine guidelines, treatments and interventions for patients with asymmetric footwear. The use of the TAS can be expanded to areas where orthotics, wedges and lifts are used to treat specific injuries and pathologies.

METHODS

Four raters, two novice and two expert, independently assessed the height and density of 10 individual shoes varying in style, size, and brand. A 6"/metric/SAE digital calliper (Wayco Equipment LTD, Auckland, New Zealand) was used to measure the height of the inner and mid/outersoles. An Asker-C Durometer (Rex Gauge Company Inc., Buffalo Grove, IL., USA) was used to measure midsole density. Each shoe was marked with five lines and two dots on the medial (Fig. 1) and lateral mid/outsole to indicate calliper and durometer placement, respectively. Each rater measured all locations three times. The total asymmetry score is calculated for the rearfoot and forefoot as the sum of the thickness and/or density differences between the medial and lateral inner and mid/outersole [1] in millimetres.

Intraclass correlation coefficients were used to assess the qualitative agreement within raters ($ICC_{2,1}$); between-raters ($ICC_{2,1}$, $ICC_{2,2}$, $ICC_{2,3}$); and between-days for one novice and one expert rater ($ICC_{3,1}$, $ICC_{3,2}$, $ICC_{3,3}$). Absolute reliability of all measurements were assessed using the standard error of measurement (SEM). Relative reliability was assessed for between-day measurements using minimal detectable differences (MDD) with 95% confidence intervals. Bland-Altman plots were used to visualise the limits of agreement and look for systematic differences between raters and between days.

RESULTS

Within-rater, between-rater and between-day ICC of all measurements were good to excellent (>0.67), with the exception of one density measurement which was poor ($ICC_{2,1}$=0.31).

The SEM for digital calliper measurements ranged from 0.10 mm to 1.82 mm for novice raters and 0.19 mm to 0.60 mm for expert raters. Durometer SEM was < 5 Asker-C units for all raters. The rearfoot TAS was measured within 0.30 mm of the true score when using the average of three measurements.

The MDD for the TAS at the rearfoot and forefoot for novice raters ranged from ($MDD_{95\%}$= 0.84–1.42 mm) and expert raters ranged from ($MDD_{95\%}$= 0.54–0.96 mm), (Table 1).

No significant differences existed between raters or between days. On the Bland-Altman plots, novice raters showed a trend of measuring small midsole heights larger than the experts, and as the sole thickened, the novices measured it smaller than the experts. However, the trend was not statistically significant.

Figure 1: Medial mid/outsole measurement locations

Table 1: Minimal detectable difference of total asymmetry scores by novice and expert raters

	$MDD_{95\%}$ Single trial	$MDD_{95\%}$ Two-trials	$MDD_{95\%}$ three-trials
Novice:			
$TAS_{rearfoot}$	1.37mm	0.84mm	0.86mm
$TAS_{forefoot}$	1.38mm	1.28mm	1.42mm
Expert:			
$TAS_{rearfoot}$	0.96mm	0.57mm	0.58mm
$TAS_{forefoot}$	0.89mm	0.92mm	0.54mm

SUMMARY

The total asymmetry score is a reliable footwear assessment tool for determining the mediolateral asymmetry at the rearfoot and forefoot of conventional running shoes. Novice raters can achieve comparable reliability when assessing the TAS by measuring each location twice. Both novice and expert raters can be 95% confident a true change has occurred when TAS is ≥ 1mm.

REFERENCES

1. Sole, C.C., et al., *Patterns of mediolateral asymmetry in worn footwear.* Footwear Science, 2014. **6**(3): p. 177-192.
2. Milford, J., Sole, C.C., Sole, G., *Plantar pressure profiles of athletes with and without ankle sprains while walking barefoot and in sport shoes.* Footwear Science, 2016. **8**(1): p. 9.

DISCLOSURE STATEMENT

The authors have no conflicts of interest to disclose.

A COMPARISON OF *IN VIVO* TIBIOTALAR AND SUBTALAR KINEMATICS IN CHRONIC ANKLE INSTABILITY PATIENTS AND ASYMPTOMATIC CONTROLS USING HIGH-SPEED DUAL FLUOROSCOPY

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INTRODUCTION

Chronic ankle instability (CAI) often results from trauma to the ankle and is associated with persistent feelings of instability, pain, frequent ankle sprains, and difficulty walking on inclined or uneven surfaces [1]. It is clinically hypothesized that deleterious angular and translational (i.e. kinematic) motion at the tibiotalar and subtalar joints causes ankle osteoarthritis (OA) in CAI patients [2, 3]. However, *in vivo* measurements of tibiotalar and subtalar motion in CAI patients are not currently available, and thus, the pathogenesis of OA in CAI patients is unknown.

Joint kinematics are often derived from motion capture techniques that track the positions of reflective markers adhered to the skin over bony landmarks. However, skin-marker motion capture cannot distinguish between motion of the tibiotalar and subtalar joints as there are no palpable landmarks for placement of a skin marker about the talus. Dual fluoroscopy (DF) is an imaging modality that accurately measures three-dimensional *in vivo* bone movement and can be used to identify the independent roles of the tibiotalar and subtalar joints. The purpose of this study was to use DF to evaluate and compare *in vivo* tibiotalar and subtalar joint kinematics between CAI patients and asymptomatic controls during activities of daily living.

CLINICAL SIGNIFICANCE

Measurements of *in vivo* tibiotalar and subtalar kinematics are not available for CAI patients. These data could clarify the pathomechanical characteristics of this condition and illuminate the steps required to refine current diagnosis and treatment strategies.

METHODS

We used dual fluoroscopy to measure *in vivo* tibiotalar and subtalar joint kinematics in CAI patients during a single-leg, balanced heel-rise activity and during treadmill gait at 0.5m/s and 1.0 m/s. Data were compared to healthy controls.

RESULTS

During balanced heel-rise, 70%, 58% and 65% of the measured CAI tibiotalar internal/external rotation (IR/ER), subtalar inversion/eversion (In/Ev) and subtalar IR/ER, respectively, fell outside the 95% confidence interval (CI) of the control subjects. During 0.5 m/s gait, 50% and 60% of CAI tibiotalar dorsi/plantarflexion (D/P) and subtalar IR/ER, respectively, fell outside the 95% CIs of the control subjects. During 1.0 m/s gait, 62%, 65% and 73% of CAI subtalar D/P, In/Ev and IR/ER, respectively fell outside the 95% CIs of the control subjects. Less than 50% of the CAI joint angles for the remaining activities or directions fell outside the 95% CIs of the controls.

CAI patients often exhibited less rotational and translational range of motion (ROM) in both joints throughout the various activities. Most CAI patients demonstrated consistently less In/Ev ROM for both joints, although individual ROM values were not always outside the 95% CIs of the controls. During gait, all CAI patients exhibited less tibiotalar translational ROM than the respective means of the controls and often less than the 95% CI of the controls (Fig 1).

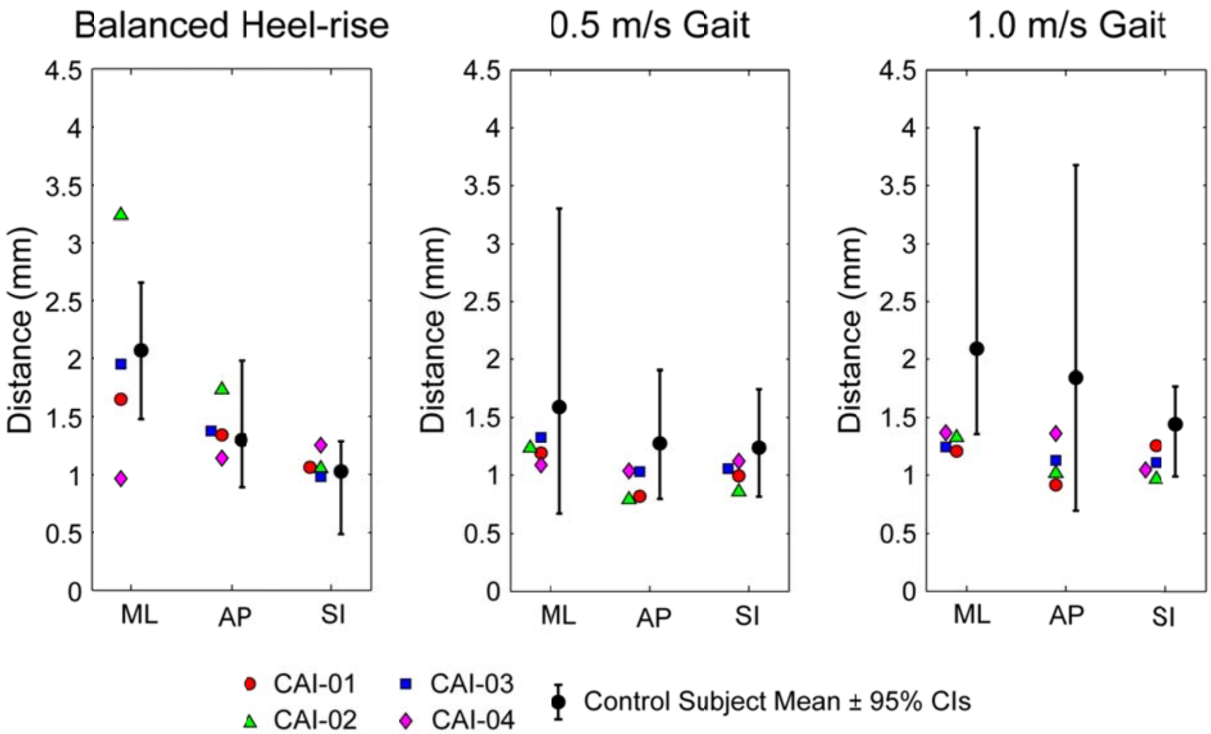

Figure 1: Tibiotalar joint translation range of motion (ROM) values for chronic ankle instability patients (colored symbols) plotted against the mean (black dots) ROM and 95% confidence interval (black bars) of asymptomatic controls. Translation directions: ML = medial-lateral; AP = anterior-posterior; SI = superior-inferior.

DISCUSSION

Results of the individual joint angles during balanced heel-rise suggest that tibiotalar external rotation, subtalar eversion, and subtalar external rotation angles were indicators of CAI, as CAI patients often exhibited greater joint angles with an opposing trend when compared to controls. Our results suggest that altered tibiotalar IR/ER, subtalar In/Ev and IR/ER and reduced translational ROM may be biomechanical characteristics of CAI.

REFERENCES

[1] Delahunt E, et al. (2010) Medicine and science in sports and exercise, 42:2106-21.
[2] Harrington KD. (1979) The Journal of bone and joint surgery American volume, 61:354-61.
[3] Valderrabano V, et al. (2009) Clinical orthopaedics and related research, 467:1800-6.

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DISCLOSURE STATEMENT

The authors have no conflicts of interest to disclose.

QUANTITATIVE ASSESSMENT OF WALKER USE DURING GAIT

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Introduction

Cerebral Palsy (CP) is a common motor disability for children and is caused by abnormal development or damage of the brain. Many children with CP rely on assistive devices such as walkers for ambulation due to muscle stiffness, weakness, poor balance, and poor coordination. Quantified movement analysis is often used to assess gait impairments in children with CP to assist with determining appropriate treatment interventions. Typically, movement analysis includes quantification of kinetics to determine joint loading patterns. However, when a walker is used for support during gait, it is difficult to interpret the kinetic data without understanding the load-bearing through the upper extremities. There is a need for an easy to use tool to accurately collect force data that indicate the loads on the walker that occur through the right and left upper extremities (UE).

Clinical significance

There is currently no simple method to quantify the amount of UE loading an individual is applying to their walker. As a result, assessment of an individual's dependence on a walker is subjective. There is also no simple method to quantitatively assess the symmetry of loading on right versus left UE.

Methods

Device Overview

The device (*InvisiForce*) is a lightweight and reusable surrogate handle that can be attached to most standard walkers. It detects the vertical force applied, which acts as an indicator of the patient's reliance on their walker.

Principle of Operation

InvisiForce detects forces using a single-point load cell (HBM – PW6d/40kg, Marlboro, MA) suspended below the walker handle. The device is oriented as a cantilever, where one side is clamped to the walker crossbar and the other side suspends a steel tube that floats concentrically around the walker handle (acting as a surrogate handle). The vertical force (Fig. 1) causes minute deflections that are measured by the load cell in terms of an output voltage. The voltage is amplified, recorded with an analog to digital converter, and normalized to the subject's body weight. The *InvisiForce* is attached to both the right and left walker handles to allow clinicians to assess vertical force through each handle individually.

Figure 1: Photo of *InvisiForce*. The red arrow indicates location of vertical force applied by the surrogate handle on the load cell cantilever.

Demonstration

The subject is a seven year old male with a diagnosis of spastic diplegic cerebral palsy and a surgical history of bilateral adductor tenotomies at age three. He uses bilateral solid ankle foot orthoses and requires the use of a reverse walker for ambulation.

The subject was referred for a clinical gait analysis study. He volunteered, with parental consent, to be tested with the *InvisiForce* on his personal walker. Testing was completed barefoot as well with shoes and braces.

The subject is a Level III in the Gross Motor Function Classification System (GMFCS). Developmental milestones were delayed as he sat at twenty-four months and began walking between the ages of three and four years.

Lower extremity and trunk motion were recorded with 28 retroreflective motion markers using 12 Eagle-4 digital cameras. Cortex 6.2 (Motion Analysis Corporation, Santa Rosa, CA) captured the motion and the analog data (voltage) from the *InvisiForce*. The data were processed and plotted using Visual3D v5 (C-motion, Germantown, MD).

Summary

Force data from the *InvisiForce* are presented in Figure 2. All data came from one representative bare walk and one representative braced walk trial. In Fig. 2a and 2b, the red and blue lines parallel to the x-axis (near the bottom of the plots) indicate right and left stance phase respectively.

In Fig. 2a, barefoot condition, loading and unloading of the right and left walker handles occur at the same time with an increased load during right stance phase. However, in the braced condition, Fig. 2b, loading becomes reciprocal. The load through the walker handles occurs during contralateral stance phase.

Fig. 2c is braced vs bare condition. Left and right vertical UE force data have been added to calculate total vertical force exerted on the walker. The braced condition has less total force through the walker handles than the barefoot condition.

Conclusion

The *InvisiForce* system placed on a subject's personal walker provides quantified assessment of the right and left UE loads on the walker. This provides the clinician with valuable insight as to how the subject is utilizing the walker for support during gait.

References

1. Alwan, M. et al. (2006) Medical engineering & Physics, 29(3): 380-389.
2. Fast, A. et al. (1995) Arch Phys Med Rehabil, 76(5): 484-491.

Acknowledgments

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Disclosure Statement

The authors have no conflicts of interest to disclose.

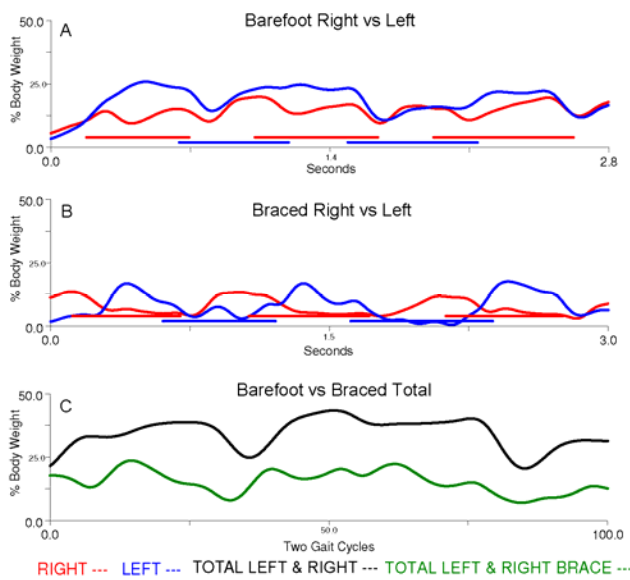

Figure 2: In Fig. 2a and 2b, the red and blue lines parallel to the x-axis (near the bottom of the plots) indicate right and left stance phase respectively.

Gait and Mobility of a Grandmother, Mother and Child with Type I Osteogenesis Imperfecta in an Urban Southeast Asian Family

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INTRODUCTION
Osteogenesis imperfecta (OI) is a highly variable, genetic disease associated with fragile bones as well as overall weakness in physical activity. Physical activity and weight-bearing situations are more difficult for those with OI, and, on average, patients with OI have decreased daily physical activity [1]. Physical activity is critical for rehabilitation early on in those with OI [2]. The purpose of this case report was to analyze 3 generations of Type I OI with respect to expressivity of the disease among the patients in terms of gait and mobility. How much familial variation – as well as differences in treatment – were both explored in the 3 generations with gait analysis and activity trackers.

PATIENT HISTORY
Three female participants with Type I OI were observed and tested at the Philippine General Hospital in Manila, Philippines: a 72-year-old, 46-year-old, and 6-year-old. For the purposes of the case report, the 72-year-old, 46-year-old, and 6-year-old will be referred to as patient ID # 1, 2, and 3, respectively. Patient ID #3 is the only patient with known history of surgical procedures: a Sofield-Millar bilateral procedure in July of 2013. None of the patients walk with assistive devices, but patient ID #2 does complain of occasional pain during ambulation. Patient ID #'s 2 and 3 have been on doses of pamidronate for less than a year and 5 years, respectively, but patient ID #1 has not taken any bisphosphonates.

CLINICAL DATA
Gillette Functional Assessment Questionnaires were filled out to identify patient concerns, goals, medical history, as well as physical functional ability. Data on height, weight, age, etc., were collected from physical examination. Only patient ID #2 complained of pain during ambulation. (See Table 1)

MOTION DATA
Gait Analysis, as well as Fitbit One activity trackers were utilized to monitor fitness/activity level in the 3 patients. Measurement of steps and steps/min were lower than healthy patients [1]. These OI patients were not as active as healthy controls, and they were below the daily recommendation of physical activity [1]. Stride length of the 3 generations – compared to normal gait data for stride length and age – was shorter than normal [3]. Patient ID #1 had the lowest amount of steps, steps/min, as well as speed of all three patients. (See Tables 1 and 2)

Table 1: Gait Parameters of Patient ID #'s 1, 2, and 3

Patient ID #	Age (yrs)	Height (cm)	Daily Steps (wkday)	Daily Steps (wknd)	Steps/Min (wkday)	Steps/Min (wknd)	Speed (m/s)	Stride Length (m)
1	72	126	4,746	3,883	3.30	2.70	.66	.79 ± .11
2	46	133	12,842	13,807	8.92	9.59	.92	.92 ± .03
3	6	93	8,043	11,827	5.59	8.21	.92	.69 ± .06

Table 2: Daily Activity Level of Patient ID #'s 1, 2, and 3

Patient ID #	Minutes Sedentary	Minutes Lightly Active	Minutes Fairly Active	Minutes Very Active
1	9210	854	0	0
2	7277	2627	142	15
3	7734	2274	49	1

TREATMENT DECISIONS AND INDICATIONS

Abnormal gait patterns within the 3 generations can be attributable to their Type I OI. Possible treatment options could include cycles of bisphosphonates, such as pamidronate, to help increase bone density. Patient ID #'s 2 and 3 have already been on doses of pamidronate, as mentioned in the above Patient History section. Cycles of bisphosphonates, as well as proper planning of rehabilitation and physical therapy treatment, have shown to aid in prevention of shorter stature and constant bone fractures in those with OI [4].

SUMMARY

One family – 3 generations – of Type I OI was observed in an Asian urban setting. The clinical assessment as well as motional analysis was typical of those with Type I OI, with abnormal gait parameters such as steps, steps/min, and stride length, compared to their healthy counterparts. This clinical case demonstrates the utilization of Gait Analysis and Fitbit One activity trackers as a basis of measurement for each patient's quantitative motional data.

DISCLOSURE STATEMENT

The authors have no conflicts of interest to disclose.

REFERENCES

1. Pouliot-Laforte et al. (2015) Physical activity in youth with osteogenesis imperfecta type I
2. Brizola et al. (2014) Muscle Strength, Joint Range of Motion, and Gait in Children and Adolescents With Osteogenesis Imperfecta
3. Sutherland et al. (1986) Normal Gait: An Introduction
4. Hasegawa et al. (2008) Growth of infants with osteogenesis impefecta treated with bisphosphonate

FOCAL DYSTONIA: A CASE STUDY

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PATIENT HISTORY

The patient presented as a 10+1 yo male with a diagnosis of left lower extremity focal dystonia of unknown origin with an unusual walking pattern. The referring physician requested guidance as to which muscle(s) to treat with botulinum toxin to improve the patient's walking and running patterns. The family was concerned about the patient's inability to keep up with friends.

CLINICAL DATA

Strength, selective control, and range of motion throughout the lower extremities were found to be within normal limits. Muscle tone in the gastrocnemius, hamstrings, and quadriceps was normal. Observational gait analysis indicated a disruption, or "catch", about the left hip during swing phase, resulting in a variable foot and knee position in terminal swing.

MOTION DATA

Prior to treatment, the patient was tested in both self-selected and fast walking conditions with motion capture based on the Newington gait model[1] and electromyography (EMG) of the rectus femoris (RF), vastus lateralis (VL), medial hamstrings (MH), gastrocnemius, and gluteus maximus (GM). (Figure 1).

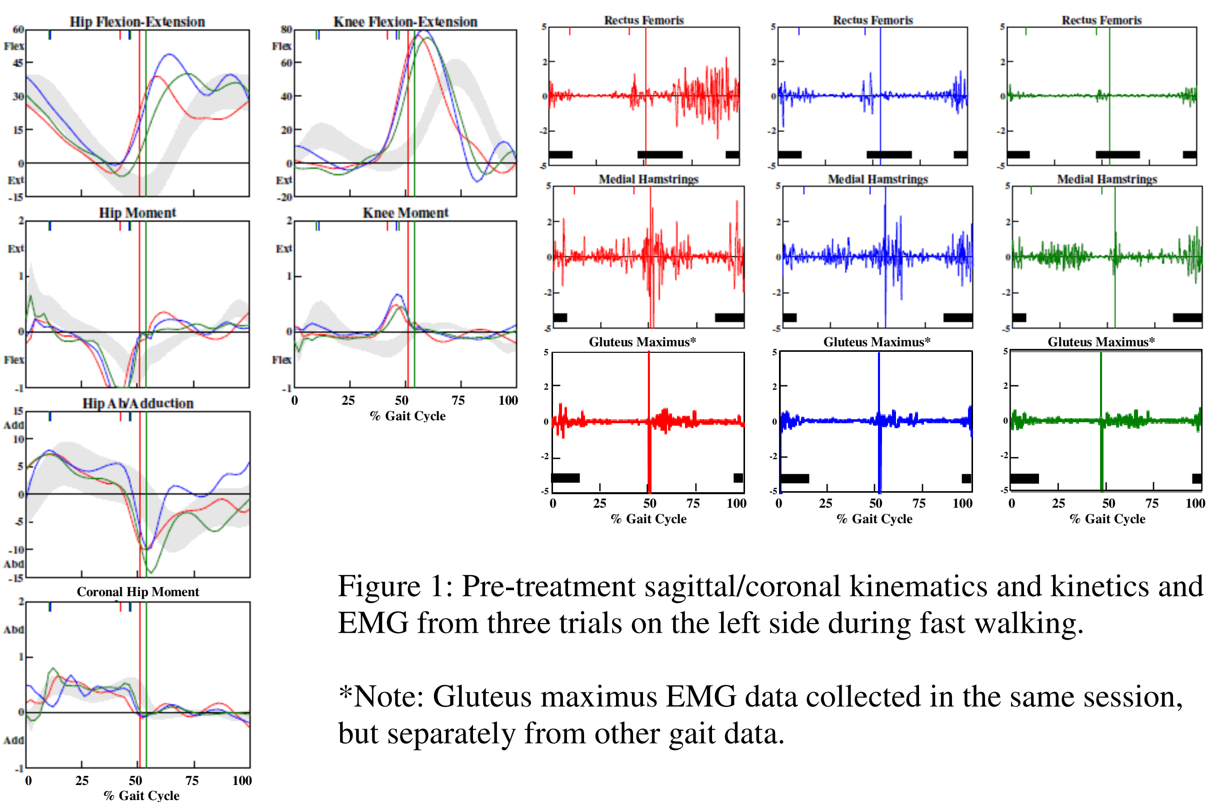

Figure 1: Pre-treatment sagittal/coronal kinematics and kinetics and EMG from three trials on the left side during fast walking.

*Note: Gluteus maximus EMG data collected in the same session, but separately from other gait data.

TREATMENT DECISIONS AND INDICATIONS

Kinematic disruptions about the left hip in swing along with increased hip extensor and abductor moments in swing were appreciated. Inappropriate activity of RF, VL, MH, and GM in swing was noted. Between trial variability in the hip/knee kinematics and RF/VL activity suggested RF/VL compensation to facilitate knee extension. Inappropriate MH activity simultaneously identified MH as potential disruptors of hip flexion and facilitators of knee flexion. Consistent inappropriate GM activity in initial swing implicated GM as a possible etiology for disruption of hip flexion. In order to isolate the muscle responsible for the disruption, only botulinum toxin injections to the GM were recommended.

OUTCOME

A repeat gait study was performed six weeks after botulinum toxin injections to the left GM. (Figure 2) Post-treatment data (fast walking) demonstrated improvement in walking speed (pre: 97%, post: 127%) with minimal disruptions at the hip. A single trial (red) showed continued disruption at the hip in swing, along with inappropriate activity of the medial hamstrings. During a separate trial (green), hip flexion in swing was more normalized along with nearly absent inappropriate hamstring activity. Based on these findings, it was recommended that the next round of botulinum toxin injections include both the GM and MH.

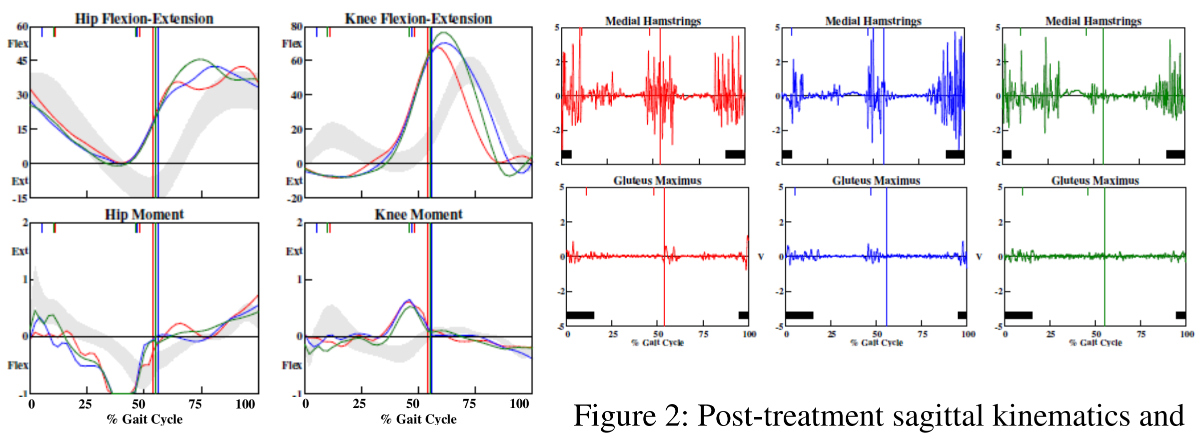

Figure 2: Post-treatment sagittal kinematics and kinetics and EMG from three repeated trials on the left side during fast walking

SUMMARY

This case illustrated the effective integration of kinematic, kinetic and EMG data for treatment planning. Also, it highlighted the use of trial-to-trial variability between individual walking trials to identify underlying primary and compensatory gait deviations. A post treatment study will be obtained to determine the effectiveness of both MH and GM botulinum toxin injections.

DISCLOSURE STATEMENT

The authors have no conflicts of interest to disclose.

REFERENCES

1. Davis RB, et al. *Human Movement Science*, 10(5):575-587, 1991.

IMPROVED HIP JOINT CENTER PREDICTION

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INTRODUCTION

In biomechanical modeling for motion analysis, the hip joint center (HJC) is used to define the proximal location of the thigh segment. This is the point about which the hip moments are calculated and the point that defines the orientation of the thigh segment, which determines hip and knee kinematics. The HJC cannot be palpated, therefore for modeling purposes; its location must be calculated. Functional methods for HJC prediction have been shown to be accurate [1], but are difficult to perform for clinical populations with neuromuscular deficits. Therefore, regression based methods are preferred; however, these methods have been shown to yield large degrees of error. Most prediction methods utilize the pelvic landmarks (i.e. the anterior and posterior superior iliac spines) to calculate the location of the HJC [2,3,4]. Excessive adipose tissue can make these landmarks difficult to locate correctly. A new regression equation utilizing leg length has been shown to improve the accuracy of locating the HJC in cadavers [5]. The Hara regression method has been shown to have considerably less error than the Bell [2] or Davis [3] methods and comparable error to the Harrington method [4]. The purpose of this study was to compare the accuracy of the HJC location calculated with the Harrington method and the Hara method. These two methods were compared to a gold standard digital full-leg coronal radiograph.

CLINICAL SIGNIFICANCE

Accurately modeling the location of the HJC is critical for data interpretation and patient care.

METHODS

Patients (5/8 female; 3/8 left knees; BMI = 29.6 ± 6) undergoing total knee arthroplasty enrolled in a larger post-operative follow-up study were used for this analysis. The pre-operative surgical knee was used for this comparison. Tibiofemoral angles, calculated using standard digital full-leg coronal radiographs were used as the 'gold standard' for comparison. An angle measurement tool (QREADS: Clinical Image Viewer, Mayo Clinic, Rochester, MN) was used to create vectors from the center of the femoral head to the midpoint of the tibial spines and from the midpoint of the tibial spines to the midpoint of the malleoli. The tibiofemoral angle was calculated as the lateral angle between the femoral and tibial vectors minus 180, where a positive angle was a varus knee and a negative angle was a valgus knee.

The motion analysis knee varus-valgus angles were collected from a static trial. Kinematic parameters were acquired with a motion analysis system utilizing ten infrared cameras (Raptor-12 cameras, Motion Analysis Corporation, Santa Rosa, CA). A set of retro-reflective, 3D markers were placed on the body of each subject as described by Kadaba, et al [6]. The 3D coordinates of the markers data were used as input to a commercial software program (Visual3D 5.02.27, C-Motion, Inc., Germantown, MD), to calculate the joint kinematics. The HJCs were defined using the regression equations developed by Harrington [4] and Hara [5]. The knee center was found on a vector directed medially from the knee marker and at a distance of one-half of the measured knee width. Similarly, the ankle center was located by a vector directed medially from the lateral malleolus marker at one-half the distance of the measured ankle width.

RESULTS

The mean error between the gold standard x-ray measurement and the motion analysis calculation for the Harrington HJC regression method was 7.61 degrees (95% CI: -0.23, 15.46) and for the Hara HJC regression method was 2.6 degrees (95% CI: -2.08, 7.27).

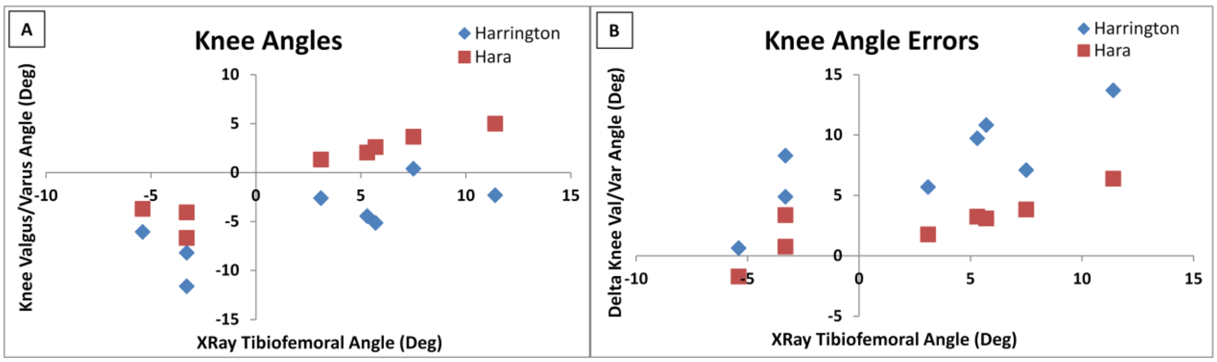

Figure 1. A. Harrington and Hara HJC knee valgus-varus angles versus x-ray tibiofemoral angles; B. Error between x-ray tibiofemoral angles and Harrington and Hara HJC angles versus x-ray tibiofemoral angles

DISCUSSION

The study demonstrated that the Hara regression method results in an improved estimate of the HJC (Fig. 1A). The HJC is both more accurate and more precise (Fig. 1B). The Hara method was based upon cadaver estimates. This study confirms that the method is reliable and valid in an in-vivo setting. These findings are limited to adults with osteoarthritis. Additional research will be needed to validate this technique in children with neuromuscular pathology.

REFERENCES

1. Piazza, S. et al. (2001) Journal of Biomechanics, 34(7):967-973.
2. Bell, A. et al. (1990) Journal of Biomechanics, 23(6):617-621.
3. Davis, R. et al. (1991) Human Movement Science, 10(5): 575-587.
4. Harrington, M. et al. (2006) Journal of Biomechanics, 40: 595-602.
5. Hara, R. et al. (2016) Scientific Reports, DOI: 10.1038/srep37707.
6. Kadaba, M. et al. (1990) Journal of Orthopedic Research, 8: 383-392.

DISCLOSURE STATEMENT

The authors have no conflicts of interest to disclose.

SOFT TISSUE ARTIFACT LEADS TO ERRORS IN BIOMECHANICAL MODEL OUTPUTS AT THE HIP JOINT

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INTRODUCTION

Biomechanical models of human motion predict metrics that cannot be measured non-invasively and *in vivo*. These models require joint angles as input, and, along with externally measured ground reaction forces, can estimate individual muscle forces and forces across a joint [1]. Errors in model inputs (joint angles) will propagate to model outputs (muscle and joint forces) and potentially lead to incorrect clinical conclusions from model predictions.

Most often retroreflective markers adhered to the skin surface serve as the basis to calculate joint angles. This is problematic because skin markers are known to suffer from soft tissue artifact (STA) [2], especially at the hip joint [3]. To our knowledge, the influence of STA on biomechanical model outputs of muscle force and joint reaction force has yet to be determined for the hip joint by comparing data from skin markers to a reference standard. We have developed a high-speed dual fluoroscopy (DF) system to measure *in vivo* bone motion, which can be used as a reference standard to calculate STA. The purposes of this study were to: 1) use DF to measure STA during walking and 2) quantify how STA influenced predictions of muscle force and joint reaction force from biomechanical models.

CLINICAL SIGNIFICANCE

The extent to which STA propagates to biomechanical model errors will facilitate interpretation of outputs used for surgical planning and studies of clinical populations.

METHODS

Eleven subjects signed informed consent to participate in this University of Utah IRB approved study (mean (SD) age: 23.2 (2.2) years, BMI: 21.1 (1.9) kg/m^2). Subjects were imaged simultaneously with a 10-camera Vicon motion capture system and high-speed dual fluoroscopy (DF), both at 100 Hz, during walking on a dual-belt treadmill at their self-selected speed [4]. Spherical markers (14mm diameter) were placed bilaterally on the torso, pelvis, thighs, legs and feet. Model-based tracking (error < 1mm and 1°) was used to analyze the DF images and measure *in vivo* pelvic and femoral bony landmark positions [5]. A calibration cube housing custom reflective skin markers containing a metal bead was imaged with DF and Vicon cameras to relate the two coordinate systems [6]. The spatial positions of bony landmarks measured by DF were then transformed into the skin marker coordinate system.

For each subject, two subject-specific models were generated from a generic OpenSim model [7] that was scaled to match the subject's anatomy. The first model was scaled and tracked with skin marker (SM) positions ("SM Model"). The second model was similar to the first except DF bony landmark positions were substituted for the skin markers on the pelvis and femur ("DF Model"). Inverse kinematics, inverse dynamics, static optimization and joint reaction force analyses were performed in OpenSim 3.3. Hip joint moments, muscle forces, and joint reaction forces were compared between models using a paired t-test.

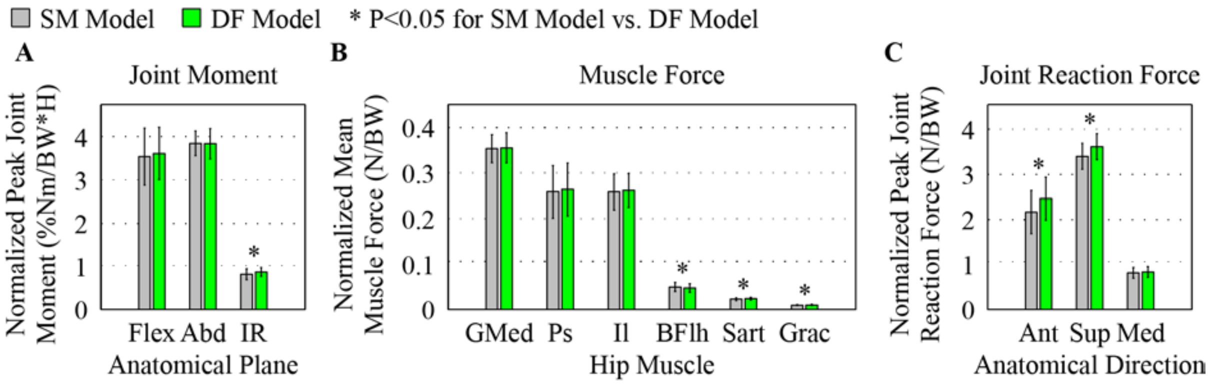

Figure 1. Comparison of joint moments (A), muscle forces (B), and joint reaction forces (C). Note that only the hip muscles with the three largest mean forces and a significant difference between the SM Model and DF Model have been shown. BW: body weight; H: height; Flex: flexion; Abd: abduction; IR: internal rotation; GMed: gluteus medius; Ps: psoas; Il: iliacus; BFlh: biceps femoris long head; Sart: sartorius; Grac: gracilis; Ant: anterior; Sup: superior; Med: medial.

RESULTS

Normalized peak joint moment differed between the SM Model and DF Model in the internal rotation plane (P=0.03) but not the flexion nor abduction planes (Fig. 1A). The biceps femoris long head generated more force in the SM Model than the DF Model (P=0.03), and the sartorius and gracilis generated more force in the DF Model compared to the SM Model (P=0.01 for both) (Fig. 1B). The normalized mean muscle force did not differ between the DF Model and SM Model for the remaining muscles. The DF Model generated larger normalized peak joint reaction forces than the SM Model in two of the three anatomical directions (anterior $P<0.001$, superior P=0.01) (Fig. 1C).

DISCUSSION

To our knowledge this study represents the first assessment of the effects of STA on hip joint mechanics predicted by biomechanical models using a reference standard. The relative importance of differences will depend on the research question; however, our results suggest that models based on skin markers underestimate important hip joint metrics, including peak joint moments, muscle forces and joint forces, relative to models driven by dual fluoroscopy.

REFERENCES

1. Steele, K.M. et al. (2012) Gait & Posture 35(4):556-560.
2. Leardini, A. (2005) Gait & Posture 21(2):212-225.
3. Fiorentino, N.M. et al. (2016) *submitted* Gait & Posture.
4. Fiorentino, N.M. et al. (2016) Gait & Posture 50:246-251.
5. Kapron, A.L. et al. (2014) Journal of Applied Biomechanics 30(3):461-470.
6. Fiorentino, N.M. et al. (2016) Annals of Biomedical Engineering 44(7):2168-2180.
7. Rajagopal, A. et al. (2016) IEEE Transactions Biomedical Engineering 63(10):2068-2079.

ACKNOWLEDGMENTS

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IMPORTANCE OF SUBJECT SPECIFIC KINEMATICS AND KINETICS FOR PREDICTING CARTILAGE CONTACT MECHANICS

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INTRODUCTION

Abnormal anatomy may be the primary etiology of hip osteoarthritis (OA). Cam femoroacetabular impingement (FAI), which is a condition where the femoral head is aspherical, may be the most prevalent type of hip deformity. The aspherical femoral head may alter hip kinematics and kinetics (muscle and joint reaction forces (JRFs)). Together, abnormal kinematics and kinetics may cause damage to cartilage, leading to end-stage OA [1].

Hip kinematics in cam FAI patients have been evaluated using skin marker (SM) motion capture techniques, but results have been contradictory. Inaccuracies associated with SMs, including errors in the estimate of the hip joint center and the effects of soft-tissue artifact may make this technology ill-suited to study hip kinematics [2]. In response, we have developed and validated a dual fluoroscopy (DF) system, which quantifies hip kinematics with sub-millimeter and sub-degree errors [3]. To study hip contact mechanics, we have developed and validated a finite element (FE) modeling pipeline, which uses patient-specific representations of bone and cartilage anatomy to obtain accurate FE predictions of cartilage contact stress [4]. Historically, our models of patients with hip disorders were driven using generalized kinematics and JRFs from the literature (e.g. patients with instrumented implants [5]). However, to accurately account for differences in motion patterns of cam FAI, it may be necessary to incorporate patient-specific kinematics and kinetics into FE models, along with patient-specific anatomy.

Our ultimate goal is to understand the pathomechanics of hip OA in cam FAI patients. Toward this goal, the objective of this study was to determine the sensitivity of FE predictions to changes in kinematics and kinetics using data from SM, DF, and generalized values obtained from the literature [5].

CLINICAL SIGNIFICANCE

Our definition of the level of subject-specificity of kinematic and kinetic input necessary to represent the hip with cam FAI in FE models will clarify the roles of altered anatomy and motion patterns in the development of OA in these hips.

METHODS

One female subject with cam FAI signed informed consent to participate in this University of Utah IRB approved study (21 years of age, BMI: 24.8 kg/m^2). The subject walked at a self-selected speed (1.08 m/s) on an instrumented treadmill as hip kinematics were measured simultaneously using DF and a 10-camera Vicon motion capture system. Spherical markers were placed on bony landmarks of the lower limbs [2]. The corresponding bony landmark positions were determined from the DF data using model-based tracking [3]. Virtual DF marker and SM marker positions were combined with the ground reaction force (GRF) data in OpenSim (https://simtk.org) to calculate JRFs at the hip.

Two loading scenarios, based on the impact and active peaks of the GRF data, were evaluated for each method of defining kinematics and kinetics (generalized, SM and DF). Joint angles were applied to the femur and pelvis of the subject in Preview (https://febio.org). The sacroiliac and pubis symphysis joints were held rigid, while motion of the femur was applied in the direction of loading until the JRF was reached. Translation of the femur perpendicular to the direction of applied loading was allowed so as to reduce unrealistic stress risers caused by improper seating into the acetabulum. Cartilage and labrum contact stresses and contact areas were evaluated for the two loading scenarios using kinematic and kinetic inputs from generalized data [5], and subject-specific SM and DF kinematics and calculated JRFs.

RESULTS

For this subject, impact and active GRF peaks occurred at 15% and 47% gait, compared to 15% and 46% for the generalized data [5]. Joint angles at the impact and active peaks varied by less than 4.8° in flexion, 7.0° in abduction, and 8.8° in rotation between the three modes of data collection. For the generalized data, the JRF at the first (impact) peak was larger than the second (active) peak (1479 vs. 1295 N), while the active peak was smaller than the impact peak for both SM and DF (SM: 1115 vs. 1482 N and DF: 1172 vs. 1579 N).

The highest stresses were observed in the anterior acetabulum during the impact peak using DF model inputs (14.1 MPa). Compared to the generalized data, contact occured more anteriorly in both the SM and DF models (Fig. 1). For all three modes of kinematic and kinetic model inputs, more of the labrum was in contact during the active peak than the impact peak.

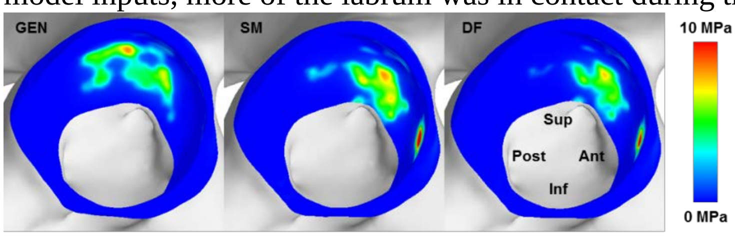

Figure 1: Contact stress during active peak for models driven using kinematics and joint reaction forces from generalized data (GEN), skin markers (SM), and dual-fluoroscopy (DF).

DISCUSSION

From this case study, the use subject-specific kinematics and calculated JRF model inputs (SM and DF) resulted in altered magnitude and location of maximum cartilage contact stresses, which is relevant to understanding the cause of OA of the hip. Further analysis is required to determine whether the differences between the subject-specificity of SM and DF model inputs are clinically relevant.

REFERENCES

1. Buckwalter, J.A. et al. (2013) Cartilage. 4(4):286-294.
2. Fiorentino, N.M. et al. (2016) Gait & Posture, 50:246-251.
3. Kapron, A.L. et al. (2014) Journal of Applied Biomechanics, 30(3):461-470.
4. Anderson, A.E. et al. (2010) Journal of Biomechanics, 43(7):1351-1357.
5. Bergmann, G. et al. (2001) Journal of Biomechanics, 34(7): 859-871.

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DISCLOSURE STATEMENT

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EFFECTS OF COGNITIVE TASKS AND DEPTH PERCEPTION ON ELDERLY BALANCE ON A STABLE AND TWO PASSIVELY UNSTABLE SURFACES

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INTRODUCTION

Previous efforts have reported that impaired depth perception adversely impacts a healthy young sample's balance performance on a passively unstable surface, specifically an off-the-shelf wobble board[1]. It is unknown, though, whether such findings extrapolate to an elderly population. It is also undetermined if similar effects can be expected for less challenging wobble board tasks or for other types of passively unstable surfaces, such as foam.

Young persons' wobble board balance performance has been worsened by concurrent quantitative and language cognitive tasks[2]. However, cognitive task effects on upper limb strategies were more substantial, as indicated by markedly increased upper limb angular velocity means[2]. The potential impact, though, of concurrent cognitive tasks on elderly balance performance and strategies, in conjunction with different passively unstable surfaces remains unknown. Subsequently, this project's purpose was to investigate the effects of impaired depth perception and cognitive task execution on two types of passively unstable surfaces. Effects on stable surface balance performance were also assessed.

CLINICAL SIGNIFICANCE

In this study impaired depth perception did not elicit torso or upper extremity strategy changes among the elderly for either modified wobble board or foam tasks, although it impeded overall balance performance on both passively unstable surfaces. Conversely, cognitive task effects were demonstrated on upper limb angular velocities for both passively unstable surfaces, as well as for stable surface, indicating changes in balance strategies. Significant effects on balance performance metrics for each surface were not observed, suggesting that strategy changes were successful.

METHODS

Nine elderly persons (4 male, 5 female), between 55 and 70 years, participated. Each subject signed an IRB approved informed consent form. Two passively unstable surfaces were foam (placed on a force plate) and a wobble board on foam. Each subject was allowed "warm up" on the wobble board prior to testing. Subjects were assumed to have experience standing directly on foam. Three reflective markers (right, left, and rear) were placed at 90° intervals along the wobble board edges. Body markers were placed in a Helen Hayes arrangement.

For each unstable surface trial, a subject was told to maintain the wobble board as still as able or remain as still as able on foam (45 sec). Each stable surface trial involved standing still on a force plate (45 sec). Motion tracking/force plate data were recorded (100/1000 Hz). Each task type was performed with: unimpaired depth perception (UD)/no cognitive task (NC); UD/quantitative cognitive task (QT), UD/language cognitive task (LT), impaired depth perception (ID)/NC, ID/QT, ID/LT. Impaired depth involved obscuring medial portions of clear lenses to eliminate visual overlap (i.e. monovision for each eye). The cognitive and language tasks were serial sevens and verbal fluency. Task instruction based performance metrics were the 95% ellipse area for board normal unit vector projection (AREA95), and fore/aft (FA) & med/lat (ML) center of pressure (COP) standard deviation (SD). Secondary measures were mean angular velocities of trunk, pelvis, and upper & lower arm segments.

RESULTS

For wobble board, impaired depth significantly increased AREA95, while cognitive tasks increased trunk, pelvis & arm angular velocities (Fig 1). For foam, impaired depth increased FA & ML COP SD, while cognitive tasks increased trunk & arm angular velocities (Fig 2). For stable surface, mean trunk angular velocity decreased with impaired depth, while FA & ML COP SD, and trunk & upper limb angular velocities increased with cognitive tasks.

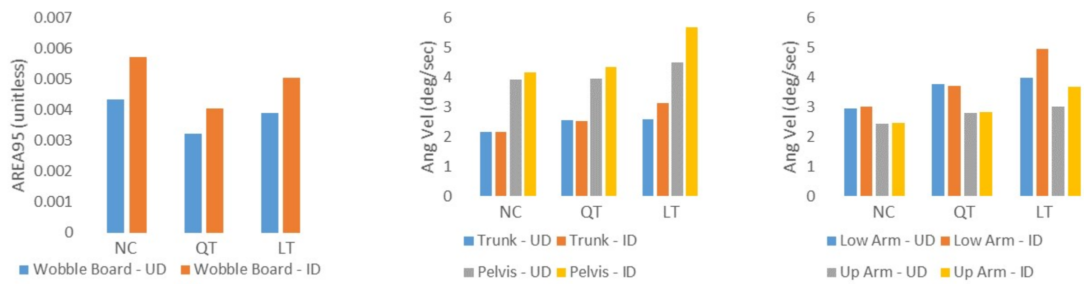

Fig 1. Balance performance and secondary measures for modified wobble board balance task.

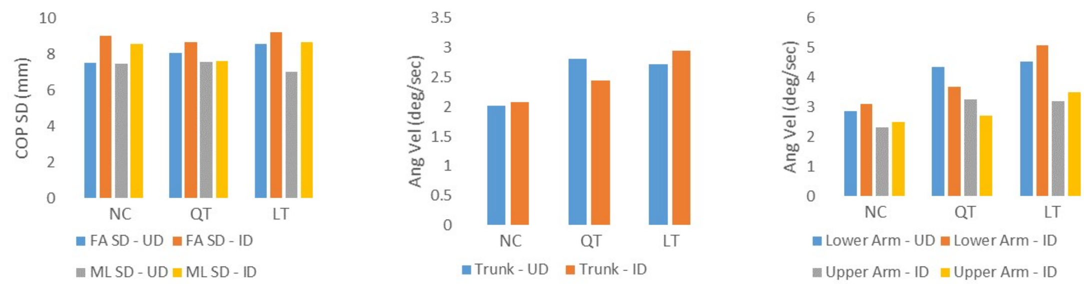

Fig 2. Balance performance and secondary measures for foam balance task.

DISCUSSION

For each surface, worsened instruction based metrics with impaired depth, in conjunction with no elevated upper body movement (particularly with no cognitive task), suggests that depth information is essential for generating the most effective lower body actuations. Conversely, consistently elevated upper body movement with cognitive tasks indicates that, with less than full attention for balance tasks, shifts towards more upper body strategies are invoked. Some level of cognitive distraction appears to allow less conscious controllers to exert influence, since degraded visual input quality wasn't sufficient to trigger notable strategy modification, even though such strategy shifts were relatively effective.

REFERENCES

1. Quesada, P.M. et al. (2011) GCMAS Annual Meeting.
2. Quesada, P.M. and Cowan, A.T. (2015) GCMAS Annual Meeting.

DISCLOSURE STATEMENT

The authors have no conflicts of interest to disclose.

Diagnostic Test for Sacroiliac Joint Pain.
–A Pilot Study to Examine Feasibility of 3D Analysis-

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INTRODUCTION

Low back pain (LBP) is a general diagnosis of motion complications in the lumbopelvic, sacral, or innominate regions that are a common reason for patients to require physical therapy. The sacroiliac joint (SIJ) is a diarthroidial synovial joint that forms a network with the lumbar spine, sacrum and the ilium bones in the pelvis[1,2]. Studies suggest that the sacroiliac joint (SIJ) may be a primary source of LBP in patients. Other studies show that hypomobility in one or both of the innominates may cause stress on the structures attached to the SIJ and also results in sacroiliac pain[2]. While the SIJ appears to be one of the underlying causes of LBP, quantifying innominate ilium motion related to the sacrum is difficult. Previously, the gold standard of quantifying this motion required radiography making it difficult to produce clinical studies. A recent study has successfully quantified this movement with a biomechanical model, yet their results (innominate angle <3°) may be difficult to reproduce in the clinic.

The March Test (Gillet Test), along with other SIJ special tests, is often used by physical therapists (PTs) and researchers to diagnose movement dysfunction at the SIJ and to evaluate the SIJ as a pain generator in LBP. To date, none of the special tests used to test for SIJ involvement have a threshold of specificity and sensitivity that is clinical useful to allow clinicians to rule in or rule out SIJ involvement. The March Test is used to assess the ilium relative to the sacrum by palpating rotational movement at the PSIS and ischial tuberosity while the subject is standing on one leg with the contralateral hip flexed to over 90 degrees; however, this test has low accuracy and reliability due to the complexity of the hip anatomy and variation between patients[1,3]. This study aimed to quantify ilium innominate movement with a reliable 3D model to quantify iliac motion with respect to the sacrum. We hypothesize that if we can quantify motion from a neutral hip to a flexed hip using motion analysis then we may be able to establish a new gold standard of measurement for SIJ dysfunction.

CLINICAL SIGNIFI ANC

The SIJ has been identified as a potential cause of LBP; however, there are currently no reliable models to assess the movement abnormalities that may cause the pain. This investigation would provide groundwork in clinical assessment of SIJ motion and dysfunctions using a 3D video capture motion analysis system. By defining axes of rotation, the movement of the ilium can be quantified with respect to the sacrum, and can ultimately be used to determine the correlation between SIJ motion and LBP. This study will use a 3D video capture to detect the different movements of the ilium among healthy individuals without hypomobility of SIJ. The results will be analyzed and compared to the recent study that used this similar biomechanical model but used a different test for SIJ pain (i.e. HABER test)[3]. We hypothesize that a test that requires deeper hip flexed position (>90°) will produce greater innominate movement of the ilium and make it easier to detect SIJ abnormalities.

HODS

Data was collected from five (n = 5) healthy young subjects. All subjects were asymptomatic of LBP and the SIJ demonstrated negative (normal) results in the March Test and Long Sit results when assessed by a PT. Each of the subjects had reflective markers placed on their shoulders, sternum, S2, PSIS, ASIS, iliac crest, and greater trochanter. Motion data was collected using a 10-camera VICON Nexus MX T-120 infrared motion capture system. Similar to the March Test, data was collected as the subject was standing, and as the subject stood on one leg with the other raised at 90° flexion (Figure 1). The subject was asked to maintain each posture for 5 seconds. The markers on the pelvis and trunk were used to calculate the Pelvis motion with using Visual 3D based on the calculation that was previously reported[3]. The innominate angle was defined as a cross-product of angles between the lines of ASIS and PSIS in each ilium (i.e. sagittal angle) while the innominate movement (defined as a transverse plane innominate angles of each right and left innominate) was calculated from the respected unit vector from PSIS. The different angles between static standing and hip flexing were used for the statistical analysis. A Paired t-test from right to left innominate was calculated to see the differences between two conditions (ilium movements of when the ipsilateral and contralateral hip was flexed).

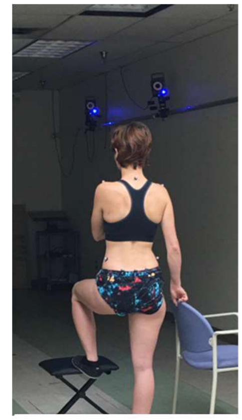

Figure 1:Data Captu ng

RESULTS

The ilium angle between right and left at static standing is approximately 2 degrees (2.17±1.53). It significantly increased when standing with passively flexed hip compared to the static standing (-6.51±3.92, p<0.05) indicating a significant movement of PSIS. There was a significant difference in the movements between the contralateral innominate and ipsilateral innominate of the tested limb in the frontal plane (p=0.03).

Table 1. Innominate Angle during One Leg Standing with Flexed Hip Position (n=5)

Angle between Both Iliums	Contralateral Innominate Movement	Ipsilateral Innominate Movement
-8.50±2.37° (p<0.01)	2.54±2.78° (p=0.018)	-1.20±2.00°* (p=0.004)

*Statistically significant compared to the contralateral innominate angle (p=0.03)

DISCUSSION

These results demonstrated greater ilium movement compared to a previous report[3]. This may indicate that a better SIJ test requires more flexed hip position than as seen in the previous report[3]. This indicates that greater hip flexed position is required for a more sensitive test to detect abnormal movement of ilium. There are some limitations in our study methods; however, this preliminary pilot provides an insight to capturing motion in the innominate. Future work will include establishing a better 3D model for further investigation of the SIJ and applying this to help select a proper clinical test for detecting SIJ symptoms and LBP.

REFERENCES

1. Vleeming, A., M. D. et al. (2012) *Journal of Anatomy,* 537–67.
2. Physical Therapy Journal." (2016). http://ptjournal.apta.org/content/79/11/1043.long.
3. Adhia, et al. (2016) *Manual Therapy* 21:100–108.

DISCLOSURE STA MENT

All authors have no conflicts of interest to disclose.

THERAPIES AND BRACING FOR THE TREATMENT OF CROUCH GAIT IN CP: A CASE STUDY

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PATIENT HISTORY

Patient is an 11 +3 year old male diagnosed with cerebral palsy spastic diplegia who was born at 28 weeks gestation. He began walking independently at 2 years. He received a posterior rollator to assist with ambulation at 10 +5 years secondary to increased crouch and frequent falling. He started wearing bilateral solid ankle foot orthoses (AFO's) at 3 years. Because of decreased tolerance, braces were changed to hinged AFO's at 11 years. He wears bilateral dynamic knee braces 2 hours daily. Patient and family concerns include decline in walking independence, endurance, and increase in knee pain.

CLINICAL DATA

Tests were completed at 7 +6, 9 +9 and 11 +3 years of age. Over time, he demonstrated decreased knee extension passive range of motion (ROM) and decreased hamstring flexibility (Table 1). His knee extension strength was graded within his available ROM.

Table 1: Summary of clinical exam and temporal findings.

	Test 1	**Test 2**	**Test 3**
BMI (kg/m^2)/Height (cm)	23.4/117	24.2/130	28.2/135
PODCI: Mobility/Pain/Global	88/89/78	79/74/66	68/56/56
Hip extension strength (knee 0 deg)	5/5	5/5	5/5
Hip extension strength (knee 90 deg)	3/3	3/3	3/3
Knee extension strength (R/L)	5/5	5/5	5/5
Ankle plantar flexion strength (R/L)	2/5	3/5	2/5
Hip Extension (deg) (R/L)	0/0	0/0	-5/-5
Knee extension (deg) (R/L)	-10/-15	-5/-5	-20/-20
Popliteal angle (deg) (R/L)	-40/-40	-45/-45	-55/-55
Ankle dorsiflexion (knee 0* deg) (R/L)	0/-5	0/0	5/5
Ankle dorsiflexion (knee 90 deg) (R/L)	5/5	10/15	15/20
Knee extension lag (deg) (R/L)	NA	-20/-20	-45/-40
Walking Velocity**- Barefoot (%)	84-92	82-85	48-55
Walking Velocity **- Braced (%)	111-115	71-86	60-66

*Knee flexion contracture; **Walking velocity % of normalized to matched leg lengths.

MOTION DATA

At test 1, motion analysis data showed a very mild crouch gait pattern with a minimal increase in knee extensor moment during terminal stance. Peak knee flexion in swing was reduced (Fig. 1). At test 2, there was increased crouch and associated increased knee extensor moment and overall reduction in knee sagittal plane range of motion. At test 3, knee sagittal plane ROM continued to reduce and the degree of crouch increased. There were similar findings bilaterally. The transverse plane profile remained within typically developing ranges across all three tests (data not shown).

TREATMENT DECISIONS AND INDICATIONS

Recommendations after test #1: Bilateral nighttime dynamic knee stretching braces for knee flexion contractures, and continue with solid AFO's for foot and ankle stability and to increase knee extension in stance. Also, physical therapy for hamstring stretching and lower extremity strengthening was recommended to improve knee extension in stance.

Recommendations after test #2: Continue physical therapy for stretching of hamstrings and gastrocnemius, and strengthening of hip extensors, abductors, quadriceps and gastrocnemius for progressive crouch position while walking. Continue solid AFO's to reduce substantially increased knee flexion in stance.

Recommendations after test #3: Bilateral distal extension osteotomies with patellar advancements and medial hamstring lengthening due to progressive crouch and significant knee flexion contracture noted bilaterally.

TREATMENT OUTCOMES

Despite initiation of recommended interventions of bracing and regular physical therapy, the comparison between tests 1 and 2 demonstrated a decline in function in terms of crouch (Fig. 1). There were similar findings between tests 2 and 3 as well as loss of independent ambulation.

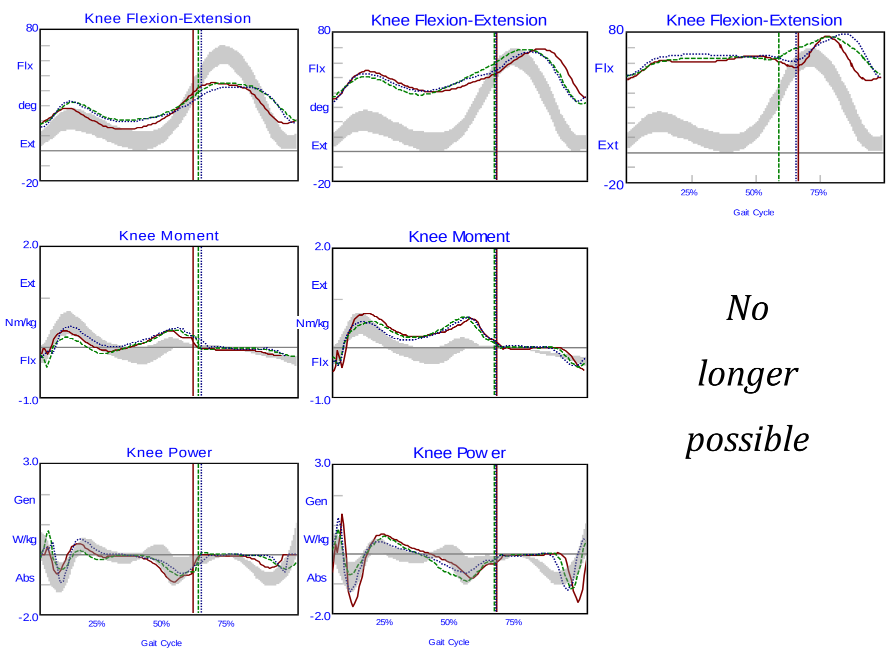

Figure 1: Comparison of multiple gait cycles for the left knee sagittal plane kinematic, moment and power for a) test 1, b) test 2 and c) test 3 (kinematic only).

SUMMARY

It is clear from the following case that although CP is a static condition, gait deviations can become more significant over time even with intensive therapies and bracing. Motion analysis provided objective documentation of patient decline in gait function and changes in associated clinical exam measures at each time point. This case example highlights the need to track patients with CP on a regular basis during growth and builds evidence to consider alternative care options such as surgical intervention when there is a measured decline in gait function. The next step will be to evaluate orthopaedic surgical treatment outcomes at approximately one year post surgery using comprehensive motion analysis techniques.

DISCLOSURE STATEMENT

The authors have no conflicts of interest to disclose.

SUB THRESHOLD VIBRATION FOR THE ENHANCEMENT OF SENSATION AND FUNCTION IN TRANSTIBIAL AMPUTEES PRELIMINARY RESULTS

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INTRODUCTION

Amputation above the foot and ankle eliminates important cutaneous and proprioceptive pathways that aid the perception of both the environment and self-motion. The overarching goal of the current work is to investigate the stochastic resonance (SR) premise 1 as a means to enhance the sensory information gained by the residual limb of individuals with a transtibial amputation. SR involves the application of a sub-threshold noise stimulus that acts to augment a weak underlying signal. This paradigm has been shown to be effective in improving sensation at the extremities and balance of older adults and individuals with peripheral neuropathy 2, .

In the current project we are exploring the effect of delivering sub-threshold vibration to the affected limb. SR traditionally employs a white noise signal. In addition to the traditional paradigm, we are investigating the effect of vibration signals with a 1/f frequency spectrum (i.e. pink noise), based on the ubiquity of this structure in healthy biological signals 4 . e hypothesize that balance and walking will improve in the presence of sub-threshold vibration. Further, we hypothesize that a 1/f-structured signal will result in greater improvements as it emulates the structure of natural physiological processes and therefore may be more instinctively accepted and integrated. In this preliminary study of quiet standing we anticipated reductions in postural sway under the SR conditions as evidence of improved control.

CLINICAL SIGNIFICANCE

This sensory deficit associated with lower extremity amputation potentially provides a significant barrier to independent locomotion, physical activity and participation. However, an effective solution is yet to be found, despite advances in prosthetic technology.

METHODS

Eight participants with a unilateral transtibial amputation were recruited from local prosthetics clinics (7M, 1F; mean(sd): 5.9(11.5)yrs, 1. 1(0.07)m, 97.5 (12.)kg, time since amputation 9. (5.2)yrs). hite noise and 1/f signals were created in Audacity (Carnegie Mellon University, Pittsburg, PA) and delivered by a commercially available tactor (Tactuator BM C, Tactile Labs Inc., Montreal) via a voice recorder (S- 00S, Olympus Corp, PA) and in-house built amplifier. The tactor was secured to the skin parallel to the upper thigh using a patch of hypoallergenic tape. The vibration was set below the threshold of perception for each participant. Levels were individually determined using a modified 4-2-1 protocol 5 . For each participant, three vibration conditions were tested - no vibration, white noise and 1/f (pink) noise, with order randomized and participants blinded to the condition of vibration. A period of at least 5 minutes was enforced between conditions to reduce potential carryover effects.

For each condition participants stood for 90 seconds with their feet as close together as possible on two adjacent parallel force plates (Optima, AMTI Inc., atertown, MA) fixating on a wall-mounted cross at eye level. The outlines of the feet were marked to maintain a consistent foot posture across trials. Centre of pressure trajectories were calculated from the combined output of the two force plates in isual D (C-motion, Bethesda, MD), and root mean square sway and sway range in anterior-posterior and medial-lateral directions extracted. A repeated measures ANO A was conducted to determine group differences between conditions.

RESULTS

Sway parameters individually showed small differences, however varying responses across participants resulted in no statistical difference across vibration conditions (Fig. 1).

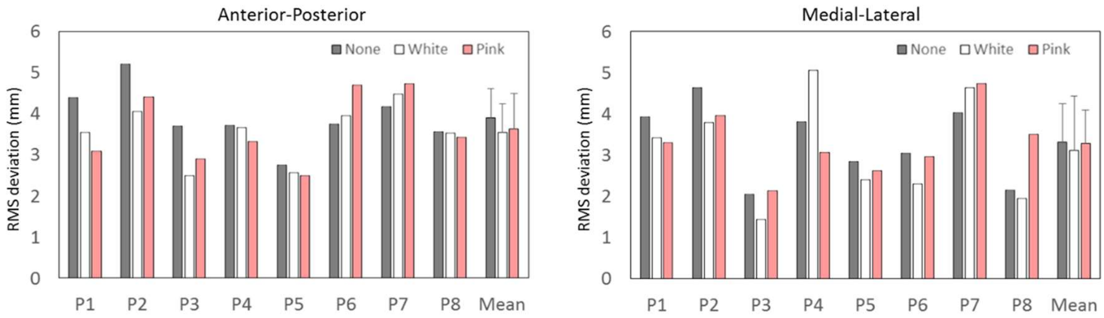

Figure 1. Root mean square (RMS) center of pressure deviation during quiet standing under conditions of subthreshold vibration none, white noise, pink noise (n ; AP: p 0.2 7, ML: p 0.55). Similar results were seen for range (AP: p 0.7 , ML p 0. 4).

DISCUSSION

Our preliminary data did not provide conclusive evidence that sub-threshold vibration has an effect on sway patterns during quiet standing. It is possible that the task of quiet standing was not sufficiently challenging to permit profound improvement, or that these traditional measures of sway were not sensitive enough to detect differences in control of posture.

The heterogeneity of the group was fairly large. Thresholds of perception of light touch and vibration stimuli were varied, and participants reported experiencing phantom pain and/or sensations to varying extents. The assessment of the response to the intervention in a larger group of participants may enable us to determine whether a subset of individuals may benefit. Further analyses will include non-linear measures and a broader range of tasks.

REFERENCES

1. Benzi R, et al. (19 1) Phys A: Math Gen 14, L45 -L457.
2. Liu , et al. (2002) Arch Phys Med Rehabil, (2), 171-17 .
. Priplata AA, et al. (200) Lancet, 2(9 90), 112 -1124.
4. Stergiou N, et al.(201) KR 5(1), 15-22.
5. Dyck, P , et al. (199) Neurology, 4 ( ), 150 -150 .

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DISCLOSURE STATEMENT

No conflicts of interest to disclose.

Effect of unilateral hip orthosis stiffness on gait in healthy individuals

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INTRODUCTION

In patients with hemiparesis, peak ankle dorsiflexion in the swing phase during gait is reduced [1]. The insufficient ankle dorsiflexion may lead to falls because ankle dorsiflexion is related to toe clearance [2]. Hip flexion has an important role in compensating for decreased ankle dorsiflexion during gait [3]. However, in patients with hemiparesis, there is decreased peak hip flexion in the swing phase during gait because of hip flexor weakness [4]. Therefore, many patients with hemiparesis require hip flexion assistance to prevent falls during gait.

To compensate for hip flexor weakness, some patients with hemiparesis wear a hip orthosis. However, the appropriate hip orthosis stiffness during gait is unclear. In a previous study, ankle foot orthosis stiffness affected the ankle angle [5]. Therefore, hip orthosis stiffness may be an important factor for determining hip angle, although there has been no study on hip orthosis stiffness. The objective of the present study was to clarify the effect of hip orthosis stiffness on hip flexion angle during gait in healthy individuals.

CLINICAL SIGNIFICANCE

Determining the effect of hip orthosis stiffness on hip flexion angle during gait may help in selection of appropriate hip orthosis stiffness to prevent falls in patients with hemiparesis.

METHODS

Ten healthy individuals participated in the present study. Inclusion criteria were no previous history of lower limb injuries and no obvious gait abnormalities. The present study was approved by the local ethics committee of Tohoku University of Medicine, Japan.

The three-dimensional coordinates of 42 reflective markers attached to 12 segments were measured using a three-dimensional motion analysis system consisting of eight cameras operating at 120 Hz. A 12-segment model based on anthropometric data was used as suggested by Dumas [6]. The participants were asked to walk at a self-selected speed along a 7-m walkway with and without a new hip orthosis (ACSIVE, Imasen Engineering Corporation, Gifu, Japan) on the unilateral side, which was evaluated under three conditions of the hip orthosis stiffness. Data analysis was performed using a custom Matlab program. We calculated representative kinematic and kinetic data during gait. These parameters were compared among each condition of hip stiffness using ANOVA.

RESULTS

There were significant differences ($p < 0.05$) in representative parameters between the four conditions of hip orthosis stiffness (Table 1).

Table 1: Gait parameters (mean ± SD) for the different conditions of hip orthosis stiffness

	No orthosis (NO)	**High stiffness (HS)**	**Middle stiffness (MS)**	**Low stiffness (LS)**
Gait speed (m/s)	1.29 ± 0.19	1.30 ± 0.16	1.30 ± 0.17	1.30 ± 0.14
Peak hip flexion angle in swing (degree)	28.0 ± 5.9[ab]	32.0 ± 4.6 [ad]	30.4 ± 4.7[b]	28.9 ± 4.7[d]
Peak hip extension angle in late stance (degree)	12.8±8.0[ab]	9.0±7.0[a]	9.8±7.2[b]	11.0±7.1
Peak knee extension moment in late stance (Nm/kg)	0.60 ± 0.17[a]	0.65 ± 0.20[a]	0.64 ± 0.19	0.64 ± 0.17
Peak knee power absorption in late stance (W/kg)	−2.57 ± 0.75[abc]	−2.90 ± 0.81[a]	−2.87 ± 0.81 [b]	−2.92 ± 0.78[c]
Peak hip power absorption in stance (W/kg)	−1.25 ± 0.49[ac]	−1.41 ± 0.52 [a]	−1.34 ± 0.50	−1.38 ± 0.49[c]
Negative work at knee joint during gait (J/kg)	-0.31±0.10[a]	-0.35±0.10[a]	-0.34±0.10	-0.33±0.09
Negative work at hip joint during gait (J/kg)	-0.37±0.15 [ac]	-0.41±0.15 [a]	-0.39±0.14	-0.40±0.15 [c]

[a] $p < 0.05$ comparing NO with HS. [b] $p < 0.05$ comparing NO with MS.
[c] $p < 0.05$ comparing NO with LS. [d] $p < 0.05$ comparing HS with LS.

DISCUSSION

Our results suggest that hip orthosis stiffness affected not only hip flexion angle during gait but also energy absolution at the knee and hip joints in healthy individuals. Data demonstrate that the use of hip orthosis could compensate for hip flexion during gait, although more muscle work may need to be performed, particularly at the knee joint. This is the first study indicating the effects of hip orthosis stiffness on kinematic and kinetic data during gait in healthy individuals.

Increased hip flexion corrected by the hip orthosis may generate more external knee flexion moment in late stance phase during gait. Thus, the use of hip orthosis with increased stiffness necessitates generating more internal extension moment, knee power absorption in the late stance phase, and negative work at the knee joint. However, increased negative work at the hip joint could be affected by increased hip orthosis stiffness. Therefore, patients with knee extensor weakness should apply increased hip orthosis stiffness carefully. In future studies, we need to examine the effects of hip orthosis stiffness on kinematic and kinetic data during gait in patients with hemiparesis.

REFERENCES

1. Lamontagne, A. et al. (2002) Gait and Posture, 15(3):244-255.
2. Green, P.J. et al. (2000) Clin Biomech, 15(7):536-540.
3. Roche, N. et al. (2015) Clin Biomech, 30(3):219-225.
4. Kerrigan, D.C. et al. (1991) Am J Phys Med Rehabil, 70, 294-300.
5. Kobayashi, T. et al. (2010) Gait and Posture, 33 (4):721-723.
6. Dumas, R. et al. (2007) J Biomech, 40(3):543-553.

DISCLOSURE STATEMENT

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WALKER ASSISTED AMPUTEE GAIT ANALYSIS USING A MOBILE PROSTHESIS INTEGRATED SENSOR SYSTEM

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INTRODUCTION

More than 1.8 million people use walkers in the United States alone [1]. Those with lower extremity (LE) limb loss typically depend on assistive devices, such as walkers, to assist their stability and avoid falls while standing and moving [2]. Clinical gait analysis is more challenging in amputee walker users due to difficulties in contrasting the limb loading and walker loading from the force plate data. However, we propose a method for quantitative assessment of walker-assisted amputee mobility using a mobile prosthesis integrated sensor. The iPecs™ is a portable, lightweight device that fits into a LE prosthesis to continuously measure three-dimensional forces and moments being applied to the residual limb [3-4].

The study aim was to evaluate the feasibility of application of the iPecs™ for evaluation of walker-assisted amputee gait, and subsequent quantitative assessment of the effects of prosthetic foot componentry.

CLINICAL SIGNIFICANCE

Clinicians and researchers will be able to objectively evaluate walker-assisted amputee gait using the proposed method that integrates mobile prosthesis sensors. Ultimately, clinicians could quantitatively assess the outcomes of prosthetic componentry without the need for conventional gait analysis, which could prove to be a lower-cost alternative.

METHODS

Gait analysis was conducted on one female participant with right transtibial amputation who used an anterior walker. The subject was a community ambulator, capable of traversing common environmental obstacles (e.g. K2-K3 activity level). Gait analysis was conducted with a 15-camera Vicon motion capture system [5], and limb kinetics were acquired using a prosthetic integrated wireless 6-axis load cell (iPecs™, RTC Electronics). The subject performed multiple over-ground walking trials at normal, slow, and fast walking speeds with the participant's current prosthesis (K3 foot: Catalyst foot) and repeated with an alternative prosthetic foot (K2 foot: Seattle Light foot) following a one-week acclimation period.

Full body kinematic data were processed and analyzed using Vicon's full body Plug-in-Gait Model [5]. The prosthetic limb kinetic data were segmented into gait cycles based on threshold analysis of the vertical force data and its derivative using Matlab (MathWorks, Inc.; Natick, MA). Temporal parameters of gait were determined from the iPecs™ data (Table 1).

DEMONSTRATION

Seven to nine gait cycles from a single trial were averaged. The prosthetic axial loading and anterior-posterior (Ant-Post) forces measured from the iPecs™ were plotted (Figure 1) to

Table 1: Gait temporal parameters (mean± SD), computed from kinetic data.

	K2 Prosthetic Foot			K3 Prosthetic Foot		
	Slow	**Normal**	**Fast**	**Slow**	**Normal**	**Fast**
Gait Cycle Duration (sec)	1.65 ± 0.07	1.47 ± 0.06	1.27 ± 0.12	1.54± 0.02	1.29 ± 0.03	1.04± 0.02
Stance Phase (% gait cycle)	77.5 ± 5	74.1 ± 5	74 ± 8	76.5 ± 2	74.4 ± 4	73 ± 2

investigate the effects of self-selected walking speed (slow, normal, and fast) and prosthetic foot componentry (K2 foot, K3 foot). The main effect of speed for both K2 and K3 feet (Figure 1) was an increase in the first axial loading peak and an increase in the posterior force peak (braking peak). The temporal parameters (Table 1) show a notable increase for normal and fast walking speed for the K3 foot in comparison to the K2, while the stance phase duration is similar. These results may reflect improved balance due to shorter double stance duration for the K3 foot [6]. These findings are further supported by the full body kinematic data [5].

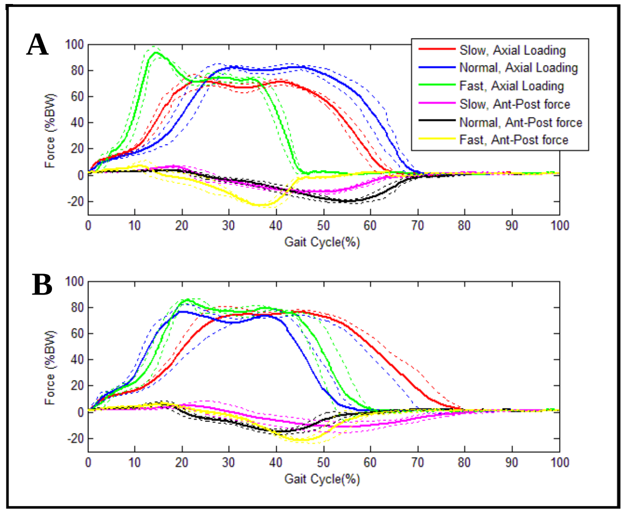

Figure 1: Mean Axial and Ant-Post prosthetic loading as a function of gait cycle (±SD) for the K3 (A) and K2 (B) prosthetic feet.

SUMMARY

We demonstrated that the iPecs is successful in detecting differences in temporal gait parameters and prosthetic loading during walker-assisted walking at varying speeds. The proposed method may prove to be a useful, lower cost clinical tool for prosthetists as compared to conventional gait analysis. Further work is underway investigating the use of the iPecs™ during ramp and stair climbing tasks, as well as over various terrain in an outdoor environment.

REFERENCES

1. Kaye, H. et al. (2000) Mobility device use in the United States. Vol. 14.
2. Bateni, H. et al. (2005) Archives of physical medicine and rehabilitation, 86(1), 134-145.
3. Koehler, S. R. et al. (2014) Journal of biomechanics, 47(6), 1542-1547.
4. Fiedler, G. et al. (2014) Journal of applied biomechanics, 30(3), 431.
5. Kopf, M. et al. (2016) RESNA/NCART 2016 Conference.
6. Murray, M. P et al. (1969) Journal of Gerontology, 24(2), 169-178.

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DISCLOSURE STATEMENT

The authors have no conflicts of interest to disclose.

EFFECTS OF STUDENT-LED BALANCE TRAINING ON A COMMUNITY-DWELLING OLDER ADULT WITH RECURRENT FALLS: A CASE STUDY.

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INTRODUCTION

Falls are a serious public health problem affecting elderly adults. Fall related injuries are associated with mortality and considerable morbidity [1]. Exercises comprising balance training and strength training have proven the most effective in reducing fall incidents [2].

Individualized exercise designed by physical therapies was found to improve balance and mobility function and reduce the risk of falls among community-dwelling older adults with a history of falling [3, 4]. However, physical therapy intervention can be expensive and often ends after a certain number of patient visits as dictated by insurance companies. Research has indicated that balance programs must be ongoing to achieve and maintain long-term fall prevention [5]. Unfortunately insurance providers do not reimburse for such preventative programs. Therefore, exploring other training programs that are more affordable and sustainable is important. The purpose of this study is to determine if a student led physical therapy program can effectively improve balance in an older adult with recurrent falls.

CLINICAL SIGNIFICANCE

This study provided valuable information that incorporates both clinical and biomechanical parameters to demonstrate the effectiveness of a student-led balance training program. In this study, biomechanical analysis was performed on the Sit-to Walk (STW) phase of the Timed Up and GO (TUG) test, which is a commonly used fall risk screening tool in clinical settings and includes STW phase. A better understanding of center of mass control during STW can provide insights for assessing mobility impairment and developing interventions for elderly fallers.

METHODS

One elderly participant, who had experienced multiple falls in the year prior to study participation, volunteered for the study. The participant is a 66 year-old, female elderly adult who resides in the community.

A biomechanical balance parameter, center of mass (COM)-ankle angle and several clinical measurements were used to assess the static and dynamic balance and mobility of the participant during the pre- and post-training evaluations. Whole body motion was captured with a six camera Vicon motion analysis system (Oxyford, UK). Inclination angle of the line formed by the COM and lateral ankle marker of supporting limb was computed for each frame during Sit-to Walk portion of the Timed Up and Go test [6]. Elderly adults with balance impairment were found to have a smaller posterior COM-ankle angle at seat-off as compared with healthy elderly adults [7]. The clinical measurements included in this study were: the Activities-specific Balance Confidence Scale (ABC), Fullerton Advanced Balance (FAB) Scale, and Dynamic Gait Index (DGI).

The participant received balance training provided by a group of four student physical therapists three times per week for four weeks. Each training session was 45 minutes.

RESULTS

The participant showed improvement in all measurements except ABC. The posterior COM-ankle angle at seat-off during STW increased from 0.71 to 8.70. The FAB increased from 35 to 39. The DGI increased from 22 to 24. The ABC decreased from 93.75 to 83.75.

DISCUSSION

The participant demonstrated an increase in posterior COM-ankle angle at seat off during STW. The farther the COM is located behind the base of support reveals an advanced level of control for a larger forward momentum at seat-off during STW. The increases in the FAB and DGI showed improved balance control after training. The ABC is a questionnaire designed to measure balance perception in the elderly. A higher number indicates better confidence in balance. The decrease in the ABC in this study seems to contradict the improvement of the other balance measurements. However, the participant indicated that she rates herself lower after training because she is more aware of her own balance limitation. This self-awareness gained from the training may reduce her future fall risk.

The outcomes of this case study suggest that student led balance training may be a cost effective intervention to alter movement strategy and improve balance for patient with fall history. More randomized control trials with larger sample size are needed in the future to strengthen the study results.

REFERENCES

1. Alexander BH. et al. (1992) Am J Public Health, 82:1020-3.
2. Robertson MC. et al. (2002) J Am Geriatr Soc, 50:905-11.
3. Nematollahi, A. et al. (2016). *Journal of aging and physical activity*, 24(2):189-95
4. Hiyamizu, M. et al. (2012) *Clinical rehabilitation*, *26*(1): 58-67.
5. Sherrington C. et al. (2011) N S W Public Health Bull, 22:78-83.
6. Chen and Chou. (2010) Gait & Posture, 31(3): 391-3.
7. Chen and Chou. (2013) Gait & Posture, 38(4): 696–701

DISCLOSURE STATEMENT

The authors have no conflicts of interest to disclose.

THE FEASIBILITY AND EFFICACY OF A STANDING SERIAL REACTION TIME TASK TO MEASURE POSTURAL MOTOR LEARNING

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INTRODUCTION

Motor learning is defined as a set of internal processes leading to a relatively permanent change in the capability for a motor skill [1]. Motor sequence learning has been investigated using a variety of experimental paradigms, including serial reaction time tasks (SRTT). For a task to test motor learning, the subject must initially be naïve to the task, and practicing the task must lead to improved performance on that task. In a SRTT, subjects are asked to map a visuospatial stimulus to a corresponding response key [2]. The most common SRTT used consists of upper extremity reaching and pointing while in a seated position. To our knowledge, no studies have used a SRTT that incorporates standing postural stepping, which could provide important insights into fall risk. Therefore, the purpose of this study was to test the feasibility and efficacy of a standing postural stepping SRTT for measuring postural motor learning. We hypothesized that, in a population of healthy young adults, there would be a dose-response relationship between the amount of practice and reaction time performance improvement and learning (efficacy) without subjects experiencing adverse events (feasibility).

CLINICAL SIGNIFICANCE

People with postural control deficits can use physical therapy services to learn or relearn motor skills to improve their balance and fall risk. While research using upper extremity SRTTs provides good insight into upper extremity motor learning, there is a void when it comes to SRTTs used to assess postural motor learning. By developing a postural stepping task that can help us better understand how practice affects postural motor learning, we will gain insights that can lead to improved recovery for people with postural deficits.

METHODS

The task was a standing postural stepping SRTT in which subjects stood on an instrumented mat, were presented with a stimulus on a computer screen, and stepped to the corresponding location on the mat as fast as safely able. Each trial entailed two consecutive 12-step sequences, with one of the sequences being random and one being repeated; subjects were blinded to the presence of the imbedded repeated sequence. Response time was collected and was defined as the time from stimulus presentation to foot touching down on the target. After completing their assigned practice dose, subjects had two days of no practice before returning for a retention test consisting of three trials of the task. *Baseline performance* was defined as mean response time on the first three trials of the first day of training. *End of acquisition performance* was defined as mean response time on the final three trials of the last day of training. *Retention performance* was defined as mean response time during retention testing. Feasibility was measured as the percentage of subjects who completed the assigned practice dose without adverse events (e.g. injury/fall). Efficacy was defined as random sequence improvement and repeated sequence improvement in both groups (via paired t-tests), as well as better random sequence learning and repeated sequence learning in the HD group compared to the LD group (via independent t-tests). These outcomes are calculated as follows: *random sequence improvement* (end of acquisition performance on the random sequences minus baseline performance on the random sequences); *repeated sequence improvement* (end of acquisition performance on the repeated sequence minus baseline performance on the repeated sequence); *random sequence learning* (retention performance on the random sequences minus baseline performance on the random sequences); *repeated sequence learning* (retention performance on the repeated sequence minus baseline performance on the repeated sequence). To test whether these expected improvements vary with dose (i.e., a dose-response), we randomly assigned subjects to a *low dose* (LD) group (1 block of 6 trials of task practice over one training day) or a *high dose* (HD) group (30 blocks of 6 trials of task practice over five training days).

RESULTS

Twelve healthy adults (8 female, mean age 29.9 years) participated. FEASIBILITY: All subjects (100%) in both groups completed all training sessions without adverse events. EFFICACY: See Table 1.

Table 1: Efficacy of the standing postural SRTT (i.e., performance improvement and learning outcomes). The asterisks (*) indicate statistically significant differences.

Outcome	**Group**	**Baseline**	**End of Acquisition**	**p-value**
Random Sequence Performance Improvement (sec)	HD	.950 (.091)	.823 (.116)	.030*
	LD	.889 (.034)	.860 (.089)	.393
Repeated Sequence Performance Improvement (sec)	HD	1.015 (.130)	.738 (.082)	<.001*
	LD	.895 (.071)	.891 (.059)	.931

Outcome	**HD Group** (n=7)	**LD Group** (n=5)	**p-value**
Random Sequence Learning (sec)	-.142 (.101)	.020 (.099)	.022*
Repeated Sequence Learning (sec)	-.217 (.136)	-.023 (.073)	.011*

DISCUSSION

These results demonstrate that this standing postural stepping SRTT was feasible for healthy young adults. Efficacy was demonstrated by the dose-response relationship, with high dose training resulting in better performance improvement and learning of the task than low dose training. Future research is needed to determine optimal practice dose for older and impaired populations, and to investigate transfer to functional tasks.

REFERENCES

1. Schmidt, R. et al. (2011) Motor control and learning: a behavioral emphasis. 5^{th} ed.
2. Boyd, L. et al. (2006) Journal of Neurologic Physical Therapy, 30(2): 46-57.

DISCLOSURE STATEMENT

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EFFECTS OF A NOVEL TWO-PHASE REHABILITATION PROGRAM ON POSTURAL CONTROL IN OLDER ADULTS

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INTRODUCTION

Falls are a major source of morbidity and disability in the aging population. Twenty to thirty percent of older adults who fall suffer moderate to severe injuries such as lacerations, hip fractures, and head traumas [1]. A serious component of falling often overlooked in rehabilitation interventions is the fear of falling. The fear of falling is part of a debilitating spiral that leads to decreased activity, loss of independence and muscle weakness [2]. The goal of this investigation was to determine if a novel two-phase rehabilitation program designed to reduce the fear of falling and increase lower-extremity muscle strength could improve postural control during falls in older adults with balance impairments.

CLINICAL SIGNIFICANCE

Results of this pilot program have the potential to transform geriatric rehabilitation practice for fall prevention.

METHODS

Phase I: Older adults participated in 8 cognitive restructuring workshops entitled A Matter of Balance (AMOB): 2hours/week, total of 16 hours, designed to restructure thought patterns relative to falls and reduce the fear of falling. Within 1-2 weeks of completion, participants enrolled in Phase II: a standardized 10 week lower-extremity strengthening program. Participants performed high-intensity concentric resistance exercise on a modified seated ergometer (Eccentron, BTE Technologies) twice per week for up to 20 minutes per session. Resistance exercise was controlled as a percent of maximum on a per session basis. Fear of falling was assessed using the Activities-Specific Balance Confidence (ABC) scale. Postural control was assessed in 4 healthy older adults during reproducible falls at three phases: baseline (T0), after Phase I AMOB (T1), and after Phase II strengthening (T2). Backward-directed falls were induced by treadmill perturbations (VGait system, MotekForce Link) occurring at slow ($2m/s^2$) and fast ($5m/s^2$) belt accelerations while standing in a static position. The Center of Pressure-Center of Mass distance (COP-COM) was assessed at three time-points in the stepping response. A 3x3 (phase x time) ANOVA with repeated measures on the time factor was used in the statistical analysis. Pairwise comparisons were analyzed with a Bonferroni correction on the time factor. All analyses were performed using SPSS (version 21.0).

RESULTS

No statistically significant interaction effects were found. However, a trend toward increasing COP-COM distance occurred after each intervention phase (T1&T2) during fast treadmill perturbations (Fig. 1). The greatest increase in COP-COM distance was found at 100% of the stepping task during fast perturbations following 10 weeks of resistance training compared with baseline (21.4cm vs. 15.5cm, p=0.006). No significant differences were found in fear of falling between phases (p=0.682).

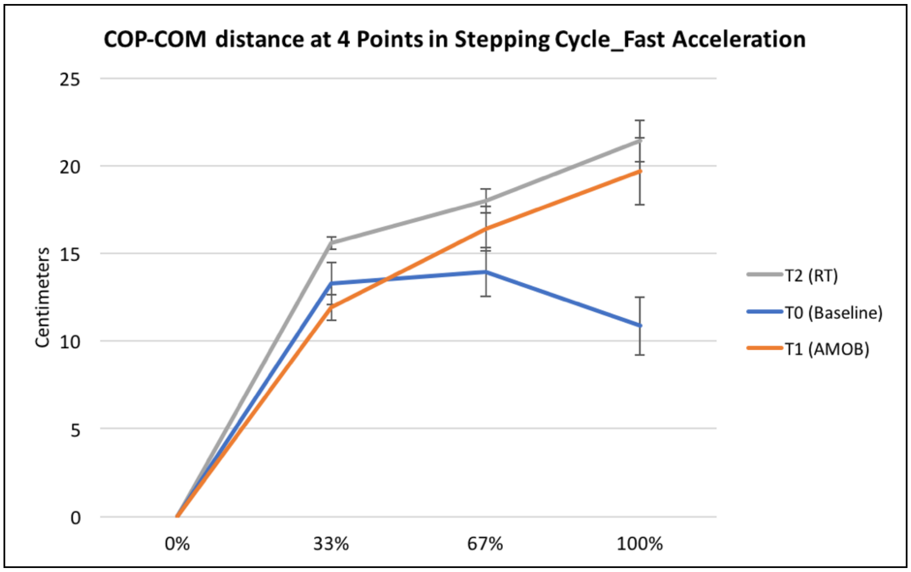

Fig1. COP-COM distance is increased following AMOB (T1) and Resistance training (T2) at 67% and 100% of the stepping cycle compared to baseline (T0) during recovery from a backward fall with high speed accelerations.

DISCUSSION

A large COP-COM distance is indicative of robust postural control [3]. A large COP-COM distance suggests the individual is able to allow straying of the COM outside of the functional base while recovering balance with a single step. Meanwhile, a small COP-COM distance represents a conservative approach to postural tasks, in that the performer does not feel stable enough to allow separation of the COP and COM [4]. These pilot data suggest that a two-phase rehabilitation program focused on cognitive restructuring and strength training can improve specific components of postural control during recovery from falls. Rehabilitation interventions aimed at reducing falls in older adults should consider adding a component of cognitive restructuring in conjunction with standard of care resistance training.

REFERENCES

1. Sterling, DA. et al. (2001) The Journal of Trauma, 50(1):116-119.
2. Vellas, BJ. (1997) Age and Ageing, 26(3):189-193.
3. Chang, H. (1999) Archives of Physical Medicine and Rehabilitation, 80, 490-494.
4. Papa, EV. (2015) Clinical Biomechanics, 30(10):1108-1113.

ACKNOWLEDGMENTS

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DISCLOSURE STATEMENT

NA

GAIT OUTCOMES OF CHILDREN WITH CLUBFEET TREATED WITH BOTULINUM TOXIN-A

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INTRODUCTION

The most common treatments for clubfoot (CF) are (i) surgical release; (ii) the Ponseti method and (iii) the French Physiotherapy (PT) method.

An essential aspect of each of these methods is the surgical lengthening of the achilles tendon (TAL). TAL is performed in 100% of surgical releases, 90% of Ponseti cases and 50% of PT cases. A corollary of TAL is that there is disruption of the gastrosoleus muscle complex, which is the major structure required for walking. This disruption results in a 20% loss of range of motion and, most importantly, a 20% loss of "power of push off" during gait, ankle strength is reduced and therefore less power is generated for forward motion) [1,2].

In the past decade, clinicians have developed an alternative, non-surgical treatment option for idiopathic clubfoot. The treatment is called MCB, for "*manipulations and casting plus adjuvant Botulinum Toxin type A (BOTOX-A) method*" [3,4]. MCB is a method based on the Ponseti technique, but replaces the surgical lengthening of the gastrosoleus complex with a BOTOX-A injection. BOTOX-A causes temporary and partial gastrosoleus muscle paralysis, permitting stretching without disruption of the Achilles tendon-gastrosoleus muscle complex, thereby preserving its integrity. MCB is the only method that does not use surgical TAL.

At BC Children's Hospital, we have used the MCB method as the primary treatment for children with clubfoot for the past 15 years. Clinical outcomes such as range of motion, reduced pain and ability to wear shoes, have been good to excellent in nearly all (90%) cases [3,4]. However, we have not yet had the opportunity to report functional outcomes (walking patterns and foot pressures) for this method. Overall, we hope to improve the lives of children with clubfoot by treating their clubfoot non-surgically to achieve an equivalent or better functional outcome than by surgical methods.

CLINICAL SIGNIFICANCE

By reporting the outcomes of the MCB method of clubfoot treatment, an alternative for the surgical lengthening of the Achilles tendon in children with clubfeet is available.

METHODS

Children with congenital clubfeet who had been treated with the MCB method and were between the ages of 5 and 6 were recruited for this study. A twelve-camera Motion Analysis system (MAC, Santa Rosa, CA) was used to record the 3-dimensional positions of reflective markers using a modified Helen Hayes marker set for both groups. Motion data were recorded for three representative walking trials on each affected foot. Data were processed using C-Motion's Visual 3D.

The study group consisted of 22 children (5.7 ±0.8 yrs), 11 with bilateral clubfeet, 3 with unilateral left CF, and 7 with unilateral right CF. This resulted in 14 left and 18 right affected feet which were compared to an age matched cohort of 14 (6.0 ±0.5 yrs). Maximum and minimum values of sagittal plane ankle angle and power were calculated and are presented.

RESULTS

Range of motion of the sagittal ankle angle during gait for the MCB method are 92% of normal and the sagittal ankle power is 80% of normal. The peak ankle dorsiflexion angle of affected right feet was significantly reduced from normal, (5° vs. 8.7°, p=0.002) however this was not the case for left affected feet, (7.2° vs 7.9°, p=0.65) (Figure 1a). There was also a reduction in peak power on the right side (1.4 W/kg vs 1.9 W/kg, p=0.02) (Figure 1b)

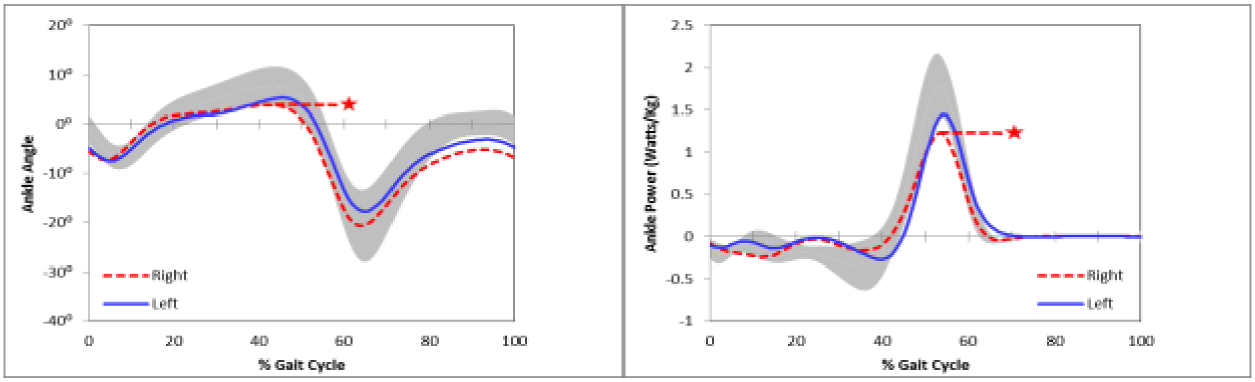

Figure1a: Average ankle angle of left and right affected feet compared to 1 sd of normal, **b)** Average ankle power of left and right affected feet compared to 1sd of normal.

DISCUSSION

The results of MCB are similar to other clubfoot interventions. Although this study showed limited reduction of dynamic range of motion at the ankles, there was a persistent loss of power, in the order of 20%, in late stance. Although statistically significant, these differences may not be considered clinically important to the patient as has been shown with patient satisfaction data. The difference with the MCB method is the absence of the TAL which can still be considered a benefit to the patient.

REFERENCES

[1] Karol, et al. *Clin Orthop Relat Res*, 2009;467 :1206-1213.
[2] Muralti et al. *J Pediatr Orthop Brit* 2005:14:206–211.
[3] Alvarez et al. *J Pediatr Orthop*, 2005;25:229–235.
[4] Alvarez, *J Pediatr Orthop*, 2009;29:570–575.

ACKNOWLEDGMENTS

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DISCLOSURE STATEMENT

The authors declare no conflict of interest.

IMPACT OF GLENOHUMERAL JOINT CONGRUITY ON SHOULDER FUNCTION IN BRACHIAL PLEXUS BIRTH PALSY

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INTRODUCTION

Approximately 1 out of every 1,000 infants will have lifelong upper extremity impairments due to brachial plexus birth palsy (BPBP) [1]. Muscle paralysis associated with BPBP injury creates an imbalance of muscle forces across the glenohumeral (GH) joint [2]. Left untreated, these imbalances lead to GH joint deformity and dysplasia [3] as well as GH range of motion deficits [4]. Tendon transfer surgeries are commonly employed to rebalance muscle forces across the joint and improve shoulder motion. Previous studies indicate that tendon transfer alone stops the advancement of GH dysplasia, but open or arthroscopic joint reduction is required to correct existing dysplasia [3]. Integrating open or arthroscopic GH joint reduction into the tendon transfer surgery increases operative and anesthesia time. Despite the frequent incorporation of joint reduction, it remains unknown whether this additional component to surgery has a measurable effect on GH contributions to global shoulder motion.

This study aims to investigate the effect of GH joint congruity on GH motion. It is hypothesized that patients who have maintained more congruent joints post-surgery will show greater GH motion than those with less congruent joints.

CLINICAL SIGNIFICANCE

A clear understanding of the importance of GH joint congruity will have an immediate and direct impact on informing clinical decision making and surgical planning. If GH joint congruity has a positive effect on shoulder function, early intervention for GH joint reduction will be indicated. If GH joint reduction does not improve shoulder function, the need to establish joint congruity during surgery would diminish, and the time under anesthesia would be reduced.

METHODS

Eleven subjects with BPBP between the ages of 4 and 17 who have undergone a latissimus dorsi and/or teres major tendon transfer surgery at Philadelphia Shriners Hospital for Children participated in this ongoing study. Subjects were grouped as having either 1) mild (n=5) or 2) moderate/severe (n=6) GH dysplasia as determined by ultrasound at the time of study participation. Motion capture (Vicon, Oxford, UK) was used to measure the 3D orientations of the trunk, scapula, and humerus segments as subjects held their arms in seven static positions (neutral, abduction, full elevation, flexion, adducted and abducted external rotation (ER), and hand-to-nape). GH joint angles at each position and displacements from neutral were calculated using a modified globe method [5] and reported as group averages. Subjects' unaffected limb data was also collected and is presented for context.

RESULTS

Both groups demonstrated greater internal rotation in the neutral position and substantial ER deficits compared to unaffected limbs (**Figure 1**). The moderate/severe group had an average of zero ER displacement for all rotation positions. The mild group achieved modest ER, but only in the abducted rotation positions. Both groups exhibited increased GH elevation in the neutral position and displayed a diminished capacity for GH elevation compared to unaffected limbs; however, the mild group possessed larger GH elevation displacements (**Figure 2**).

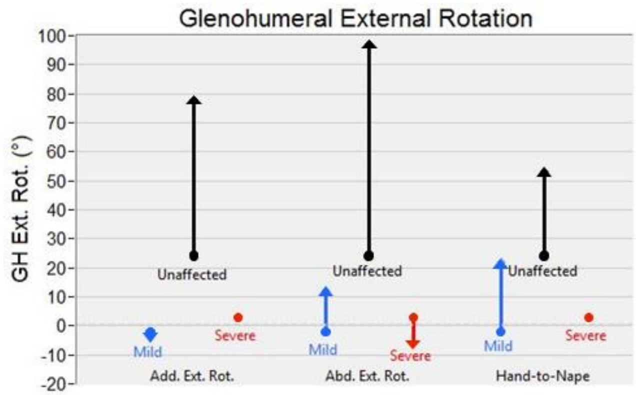

Figure 1: GH external rotation displacements from neutral. Dots show neutral orientations. Arrows show displacements from neutral to each static end position.

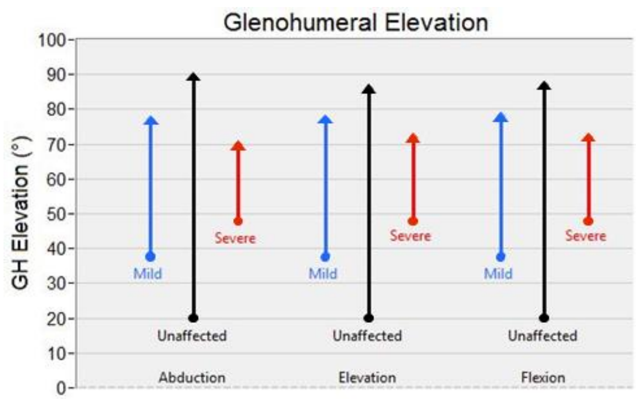

Figure 2: GH elevation displacements from neutral. Dots show neutral orientations. Arrows show displacements from neutral to each static end position.

DISCUSSION

These preliminary results suggest that a more congruent GH joint offers more GH elevation and ER motion. Rotation benefits appear to be limited to abducted positions as neither group showed GH ER displacements in the adducted rotation position. The moderate/severe group's more pronounced ER deficits may be due to larger internal rotation contractures, which are associated with GH dysplasia. Furthermore, osseous limitations such as increased glenoid and/or humeral head retroversion may alter the range of motion of the GH joint or moment arms of muscles spanning the joint, compromising ER. These results are preliminary and data collection to develop a larger sample of subjects is ongoing.

REFERENCES

1. Hogendoorn, S. et al. (2010) The Journal of Bone and Joint Surgery, 92(4):935-942.
2. Hale, H. et al. (2010) The Journal of Hand Surgery, 35(2):322-331.
3. Kozin, S. et al. (2010) Journal of Pediatric Orthopedics, 30(2):154-160.
4. Duff, S. et al. (2007) Clinical Biomechanics, 22(6):630-638.
5. Pearl, M. et al. (1992) Journal of Shoulder and Elbow Surgery, 1(2):113-118.

ACKNOWLEDGMENTS

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DISCLOSURE STATEMENT

The authors have no conflicts of interest to disclose.

Keynote #1

KINEMATIC NORMS OF WORLDTOUR CYCLISTS

WEDNESDAY, 24 May 2017
11:00A – 12:00PM

Todd M. Carver
MS Integrative Physiology
University of Colorado Boulder
Co Founder Retül University
Head of Human Performance, Specialized Bicycles
Boulder, CO

Todd M. Carver, MS is based out of Boulder, CO and works as the Head of Human Performance for Specialized Bicycles. In 2007 he co-founded the Retül Bike Fitting System that is in popular use across the industry today. Todd holds a Master of Science degree in Integrative Physiology from the University of Colorado and for the last 15 years has used his knowledge of cycling science to help professional and amateur riders improve performance and prevent injury. In his free time, Todd enjoys spending time with his wife and two boys, cycling, and skiing.

Podium Session #2

SPORTS

MODERATED BY:

Susan Kanai, PT: Center for Gait and Movement Analysis Children's Hospital Colorado, Denver, CO

Stephen Hill, PhD: Motion Analysis Research Center Samuel Merritt University, Oakland, CA

1. **Unique Balance Domains for Balance Error Scoring System (BESS) and Y-balance Tests**
   Michael Orendurff, Christopher Villarosa, Kevin Dinglasan, Kerry Peterson, Carey Hintze

2. **Quantifying Joint Coordination in Anterior Cruciate Ligament-Reconstructed Individuals**
   Kylie Davis, Brooke Sanford, John Williams, Audrey Zucker-Levin

3. **Hop Distance Symmetry Does Not Indicate Normalization of Biomechanics in Pediatric Athletes Post-ACL Reconstruction**
   Nicole Mueske, Christopher Brophy, J. Lee Pace, Curtis VandenBerg, Mia Katzel, Tracy Zaslow, Tishya Wren

4. **Biomechanical Asymmetries in Drop Jump Landing Improve During Rehabilitation Following Anterior Cruciate Ligament Reconstruction in Pediatric and Adolescent Athletes**
   Nicole MueskeJ. Lee Pace, Tracy Zaslow, Bianca Edison, Curtis VandenBerg, Mia Katzel, Tishya Wren

5. **Asymmetrical Intra-Limb Coordination at Return to Sport Following Anterior Cruciate Ligament Reconstruction**
   Adam King, Kelci Besand, Craig Garrison

6. **Landing Mechanics in Uninjured Adolescent Athletes: A Review of Symmetry in Selected Knee Kinematic and Kinetics**
   Sylvia Öunpuu, Erin Garibay, Jessica Lloyd, Nicholas Giampetruzzi, Danielle Suprenant, Matthew Milewski

7. **Comparison of Balance Control During Dual-task Walking in Adolescent Athletes After Concussion: A Multicenter Study**
   Tishya Wren, Matthew Solomito, Regina Kostyun, Yen-Hsun Wu, Nicole Mueske, Tracy Zaslow, Li-Shan Chou, Sylvia Öunpuu

8. **The Effect of Knee Support on the Knee Loads and Motions of Baseball Catchers in Deep Squats**
   Emily Dooley, James Carr, Eric Carson; Shawn Russell

UNIQUE BALANCE DOMAINS FOR BALANCE ERROR SCORING SYSTEM (BESS) AND Y-BALANCE TESTS

Michael Orendurff[1] Christopher Villarosa[2], Kevin Dinglasan[1], Kerry Peterson[1], Carey Hintze[1]
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INTRODUCTION

Balance is a critical element of mobility and functional performance, and errors in balance may contribute to injury risk and hinder locomotor success. Several effective injury prevention programs appear to improve dynamic balance and the resistance to falls from perturbations such as collisions with objects or other individuals that can result in injuries[1]. Several screening assessments, such as the Balance Error Scoring System (BESS) and the Y-Balance test have been developed, but because balance and strength variables may overlap, it is unclear whether these tests evaluate covariate gains of balance. These neuromuscular and strength differences may create unique domains of balance that are differently influenced by injury prevention programs. The purpose of this project was to evaluate the covariance relationship between two commonly used balance evaluation measures: the BESS and the Y-Balance test.

CLINICAL SIGNIFICANCE

Determining the relationship between the BESS and the Y-Balance tests may reduce the number of tests needed to determine balance improvements from interventions.

METHODS

Sixteen healthy, recreationally active adults gave informed consent to participant in this IRB-approved protocol. Over a 10-day period, each participant completed the BESS and the Y-Balance protocol four times following a standardized warm up (Figure 1). The total error score on the BESS test for both firm and foam surfaces was compared to the total distance achieved in the anterior, posteriolateral and posteriomedial directions on the Y-Balance test on each day using simple linear regression. It was hypothesized that the R^2 would be in the 0.25 to 0.50 range, indicating some overlap in balance construct domains evaluated by the Y-Balance and BESS tests.

Figure 1. Balance Error Scoring System (BESS) and Y-Balance tests.

RESULTS

Contrary to the hypothesis, the combined data over all 4 days produced an $R^2 = 0.0000013$, with a slope of -0.00007 (n = 60; $P = 0.99$; Figure 2). Individual day comparisons produced similar results (Figure 3). Day 1: $R^2 = 0.0295$; Day 2: $R^2 = 0.0290$; Day 3: $R^2 = 0.0039$; Day 4: $R^2 = 0.040$.

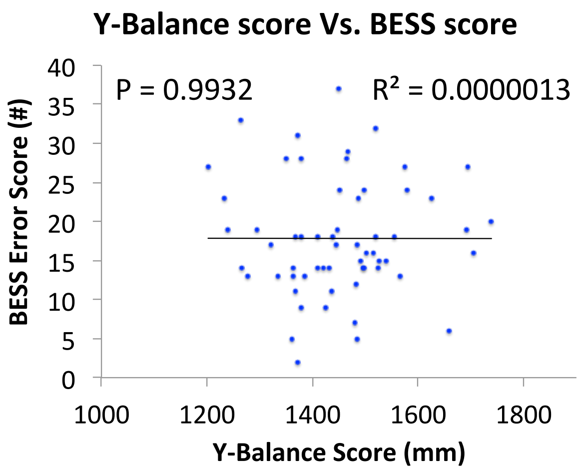

Figure 2. Y-Balance scores vs. BESS scores for 16 subjects over 4 days.

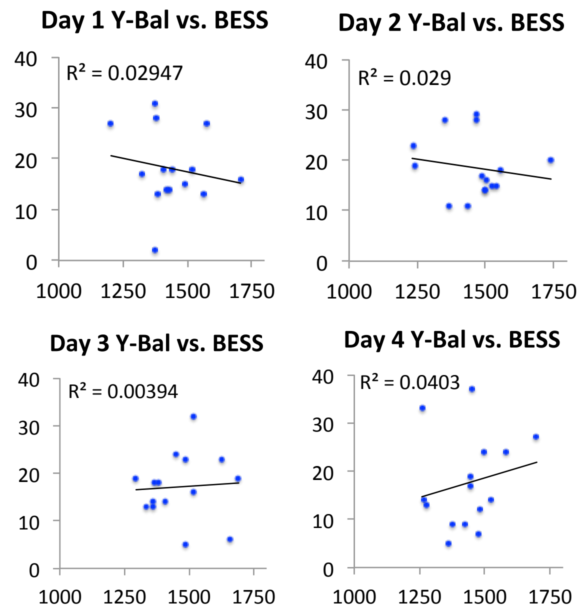

Figure 3. Y-Balance vs. BESS for each day.

DISCUSSION

These data suggest that there was no construct covariance between these two balance tests in this small cohort of recreationally active adults. The range of motion achieved on the Y-Balance test and the requirement to remain stable at the extreme end range of motion in all three directions appears to have little relationship to the BESS error scores during static standing balance on firm and foam surfaces. The Y-Balance test is likely to have a much larger strength component, especially for the knee and ankle. Greater quadriceps strength may improve performance in Y-Balance test scores but may not be a factor for the BESS test errors. These data suggest that the Y-Balance and BESS tests are unique domains of the complex construct of balance.

Dynamic balance may be improved with injury prevention programs, as demonstrated by risk reductions for both lower and upper extremity joints that often occurs as a result of collisions with other individuals[1]. These programs include drills with planned collisions with a teammate in a controlled setting, and drills designed to increase awareness of other players in close proximity. It is possible that an additional dynamic balance test may need to be developed that more closely mimics the improvements in balance and reductions in falls that appear to occur when injury prevention programs are implemented.

DISCLOSURE STATEMENT

This research was funded by a grant from E2 Technologies who had no influence on the design, conduct, data collection, interpretation or decision to analyze or publish these results.

REFERENCES

[1] Silvers-Granelli H, et al. *Am J Sports Med*; 43, 2628-37, 2015.

QUANTIFYING JOINT COORDINATION IN ANTERIOR CRUCIATE LIGAMENT-RECONSTRUCTED INDIVIDUALS

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INTRODUCTION

An estimated 200,000 anterior cruciate ligament (ACL) ruptures occur each year in the United States alone, of which approximately 100,000 are reconstructed [1]. More than 50% of ACL-reconstructed (ACLr) knees develop osteoarthritis [2] and the incidence of ipsilateral or contralateral ACL injury is about six times greater in individuals who have had ACLr [3]. Further, the Center for Disease Control and Prevention (CDC) reported an annual health care cost of more than $2 billion associated with ACL injury.

Several past studies have evaluated single joint kinetics and kinematics post ACLr, but few have analyzed these characteristics based on a dynamical systems approach. The lower extremity acts as a linked chain, which means that dysfunction at one location may elicit adaptation at another in order to maintain normal overall function. Therefore, studying the coordination between joints during a variety of tasks may provide a better understanding of ACL injury etiology as well as biomechanical changes following ACLr.

The dynamical system theory states that a healthy motor system has redundant degrees of freedom (DOF). This redundancy allows for the use of various movement patterns to achieve the same motion, and this flexibility or adaptability may be represented by a measure of coordination variability. The purpose of this study is to quantify and compare joint coordination variability in ACLr and matched uninjured individuals during gait. We have hypothesized that individuals post ACLr will show less variability in joint coordination compared to uninjured individuals.

CLINICAL SIGNIFICANCE

The quantification of joint coordination may provide clinicians with an objective measure of rehabilitation progress post ACLr and act as a testing parameter used in training and injury prevention programs.

METHODS

Following IRB approval, 20 ACLr and 20 uninjured subjects matched by age, gender, and body mass index (BMI) performed gait trials in a motion analysis laboratory. High-speed cameras (Qualysis, Gothenburg, Sweden) were used to track retroreflective markers placed on specific bony landmarks. Software (Visual3D, C-Motion, Maryland, USA) was used to create an individual biomechanical model for each subject based on anthropometrics and marker location. Lower extremity joint kinematics were calculated according to the Cardan rotation sequence XYZ, and joint coordination variability was quantified using a modified vector coding technique [4]. A coupling angle (CA) was calculated as the angle (°) relative to the right horizontal between two adjacent data points on an angle-angle plot, for which the proximal joint was plotted along the horizontal axis and the distal joint was plotted along the vertical axis. A series of joint couplings were selected for analysis based on previous research involving lower extremity pathology; hip flexion/knee flexion, knee flexion/ankle dorsiflexion, hip adduction/knee rotation, hip rotation/knee adduction, hip adduction/knee adduction, and hip rotation/knee rotation. The standard deviation (SD) of the CA at each time point across three successful trials per subject was calculated. The gait trials were shortened to only the stance phase and subsequently divided into three intervals for analysis: initial loading (0-33%), mid-stance (34-67%) and pre-swing (68-100%). The average SD was used as a measure of intra-subject variability. Subject SD values were used to obtain group averages for the ACLr and uninjured groups for comparison.

RESULTS

Individuals post ACLr showed trends of increased variability in some joint couplings: hip add./knee rot. and hip rot./knee rot. for initial loading, hip add./knee rot. and hip add./knee add. for mid-stance, and hip rot. /knee add. for pre-swing. Relevant variability measures are reported in Table 1, and Figure 1 shows an ensemble average curve of hip add./knee add. joint coupling variability for both groups.

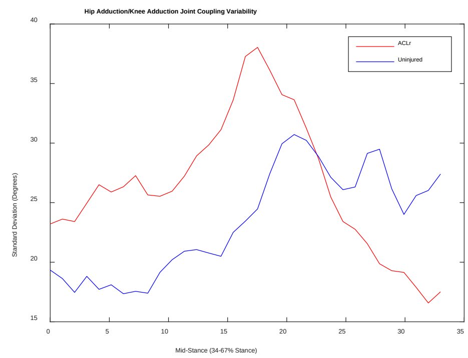

Figure 1: Hip add./knee add. variability

Table 1: Variability measures in select joint couplings

	Variability (SD in °)				
	Initial Loading		Mid-stance		Pre-swing
	Hip add/K rot	Hip rot/K rot	H add/K rot	Hip add/K add	Hip add/K rot
ACLr	18	23	26	27	17
Uninjured	15	20	24	24	14

DISCUSSION

The results showed trends of increased joint coordination variability for the ACLr group, which opposed our original hypothesis. Inspection of individual angle plots revealed that the selected stance phase intervals did not properly discriminate between gait events. Smaller intervals may be more suitable to better evaluate joint coordination in specific sub-phases of stance. Future work will include initial loading (0-20%), mid-stance (20-50%), terminal stance (50-80%), and pre-swing (80-100%) intervals. Also, characterizing motion based on proximal or distal joint dominance may supplement coordinative variability measures for a more clinically relevant evaluation. Conflicting results for coordinative variability are seen in the literature, in part due to different methodologies. Therefore, further investigation is needed to understand the implication of joint kinematics in the ACLr population.

REFERENCES

1. Evans, S. et al. (2014) International Journal of Sports Physical Therapy, 9(2): 268-273.
2. Ratzlaff, C. et al. (2010) Arthritis Research & Therapy, 12(4): 215.
3. Paterno, M.V. et al. (2014) American Journal of Sports Medicine, 42(7): 1567-1673.
4. Sparrow, W.A. et al. (1987) Journal of Motor Behavior, 19(1): 115-129.

DISCLOSURE STATEMENT

There are no conflicts of interest to disclose.

Hop Distance Symmetry Does Not Indicate Normalization of Biomechanics in Pediatric Athletes Post-ACL Reconstruction

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INTRODUCTION

Young athletes are at an increased risk for a second knee injury 7-12 months after returning to sport (RTS) following anterior cruciate ligament reconstruction (ACLR) [1]. This makes the decision of when to RTS critical and complex. RTS protocols typically include time since surgery, strength symmetry, visual movement assessment and hop for distance symmetry as criteria. However, it is unclear if suboptimal biomechanics and asymmetries persist even after hop for distance becomes symmetrical. This study assessed symmetry and biomechanics of adolescent athletes with recent ACLR during a single-leg hop for maximal distance.

CLINICAL SIGNIFICANCE

Symmetric hop distance does not indicate normal biomechanics and RTS readiness in young athletes with ACLR. Symmetry is often achieved by reducing hop distance on the contralateral side, and both symmetric and asymmetric patients offload the knee to the hip or ankle.

METHODS

39 patients with recent unilateral ACLR (62% female; age 13-18 years; 5-12 months post-surgery) and 29 controls (58% female) performed a single-leg hop for maximal distance. Three trials were collected on each leg, and the longest hop on each side was analyzed. Subjects were classified as asymmetric if hop distance on the operative limb or, for controls, the limb with the shorter hop distance, was $<90\%$ of the contralateral limb. Lower extremity 3D kinematics and kinetics between initial contact and maximum knee flexion were compared among symmetrical and asymmetrical patients' operative and non-operative (contralateral) limbs and symmetrical control limbs using ANOVA with Bonferroni post hoc tests.

RESULTS

10/29 controls (34%) and 12/39 patients (31%) were classified as asymmetric. Compared with symmetric controls, asymmetric patients hopped a shorter distance on both the operative and non-operative sides (op: 1.3 leg lengths, non-op and control: 1.6 LL, $p\leq0.04$) though there was no corresponding decrease in peak ground reaction force (GRF; op: 3.2 multiples of body weight (BW), non-op: 3.3 BW, control: 3.1 BW; $p>0.99$). Symmetric patients tended to hop a shorter distance on both sides (1.4 LL, $p=0.17$) with lower peak GRF (op and non-op: 2.8 BW; control: 3.1 BW, $p<0.10$) compared with controls.

For asymmetric patients, the primary kinematic differences compared with controls were landing more plantarflexed (op: -17.9°, control: -2.2°, $p=0.002$) with greater pelvic drop (op: -12.9°, control: -9.7°, $p=0.055$) and less knee varus (op: 0.1°, control: 2.7°, $p=0.045$). Operative limbs had lower knee flexion moments (op: 0.10 Nm/kg, control: 0.15 Nm/kg; $p=0.004$) and greater power absorption at the ankle (op: 0.47 J/kg, control: 0.29 J/kg; p=0.045), with a trend of higher dorsiflexion moments (op: 0.83 Nm/kg, control: 0.54 Nm/kg; p=0.084) and less power absorption at the knee (op: 0.71 J/kg, control: 1.1 J/kg; p=0.103).

Symmetric patients had greater peak hip flexion on both sides compared with controls (op: 70.6°, non-op: 68.3°, control: 55.2°, p≤0.001) and less varus at initial contact on the operative side (op: 0.8°, control: 2.7°, p=0.033). This resulted in higher hip flexion moments (op: 0.19 Nm/kg, non-op: 0.18 Nm/kg, control: 0.13 Nm/kg; p≤0.002), higher power absorption (op: 0.83 J/kg, non-op: 0.73 J/kg, control: 0.47 J/kg; p≤0.021), and lower knee valgus moments (op: -0.05 Nm/kg, non-op: -0.07 Nm/kg, control: -0.10 Nm/kg; p≤0.022) on both sides compared with controls, as well as lower knee flexion moments on the operative side (op: 0.10 Nm/kg, control: 0.15 Nm/kg; p<0.001).

DISCUSSION

A similar percentage of patients (31%) and controls (34%) were classified as asymmetric based on single-leg hop distance with a typical 90% threshold. This suggests that hop distance asymmetry may not reflect single leg function and RTS readiness. Furthermore, although both symmetric and asymmetric patients demonstrated biomechanical differences compared with controls, they employed different movement strategies. Asymmetric patients offloaded the knee to the ankle, while symmetric patients offloaded the knee to the hip and decreased task performance on both sides. Therefore, symmetric hop distance does not indicate normal biomechanics and RTS readiness.

REFERENCES

1. Paterno MV. et. al. (2014) Am J Sports Med, 42: 1567-73.
2. Smith, HC. et al. (2012) Sports Health, 4(1): 69-78.
3. Hewett, TW. et. al. (2010) N Am J Sports Phys Ther, 5:234-251.

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DISCLOSURE STATEMENT

One author is a consultant for Arthex and Ceterix; neither company was involved in any aspect of this study. The remaining authors have no conflicts of interest to declare.

BIOMECHANICAL ASYMMETRIES IN DROP JUMP LANDING IMPROVE DURING REHABILIATION FOLLOWING ANTERIOR CRUCIATE LIGAMENT RECONSTURCTION IN PEDIATRIC AND ADOLESCENT ATHLETES

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INTRODUCTION

Anterior cruciate ligament (ACL) injury and reconstruction (ACLR) are increasingly common in pediatric and adolescent athletes[1]. Moreover, the risk of future knee injury as well as knee osteoarthritis increases after the first incidence of knee injury[2]. The decision of when to return an athlete to sport (RTS) following ACLR is complex and usually includes consideration of time since surgery, strength and visual assessment of motion. Motion analysis during dynamic sports maneuvers can offer additional objective insight about an athlete's movement biomechanics and RTS readiness. This study examined changes in three-dimensional (3D) movement patterns and asymmetry during vertical drop jump landing between early and late stages of rehabilitation in pediatric and adolescent athletes who underwent ACLR.

CLINICAL SIGNIFICANCE

Motion analysis can augment the current clinical assessment of RTS readiness after ACLR by quantifying resolution of a knee avoidance mechanism and improvements in proximal control. Although knee loading increases from early to late rehabilitation, it remains deficient 6-10 months after surgery indicating that additional rehabilitation is needed.

METHODS

21 pediatric athletes (14 female, 7 male; mean age 15.3 yr, SD 1.9, range 10.7-17.7) who had unilateral ACLR for a non-contact ACL injury underwent motion analysis testing between 3-6 months and again 6-10 months post-operatively. 3D kinematics and kinetics were analyzed during the landing phase (initial foot contact to peak knee flexion) of a 41 cm vertical drop jump. Differences in biomechanical measures were compared between visits and between operative and contralateral limbs using paired t-tests.

RESULTS

At the first visit, the operative side exhibited less knee flexion and ankle dorsiflexion, lower vertical ground reaction force (vGRF), and lower knee and ankle sagittal external moments and power absorption compared with the contralateral side (Table1). Between visits, hip and knee flexion increased on both sides, as well as hip flexion moments and power absorption (Table 2). Power absorption at the knee increased on the operative side, with a trend of increased vGRF. Nevertheless, vGRF, ankle dorsiflexion, and knee and ankle sagittal moments and power absorption remained lower than the contralateral side at the second visit. Only minor changes between visits were observed in the frontal and transverse planes, and no asymmetry was observed in these planes at either visit.

Table 1: Kinematics and kinetics during the vertical drop jump (mean ± SD).

	Visit 1 (3-6 months post-op)			Visit 2 (6-10 months post-op)		
	Op	**Non-op**	**P**	**Op**	**Non-op**	**P**
Peak hip flexion (degrees)	92.9 ± 12.5	91.9 ± 9.8	0.30	103.3 ± 13.0	102.3 ± 12.9	0.34
Peak knee flexion(degrees)	**93.9 ± 13.8**	**97.6 ± 14.2**	**0.002**	103.4 ± 13.4	105.2 ± 13.2	0.07
Peak ankle dorsiflexion (degrees)	**27.5 ± 7.2**	**30.2 ± 7.3**	**0.02**	**26.8 ± 5.6**	**29.2 ± 5.3**	**0.02**
Peak vertical GRF (multiples of body weight)	**1.5 ± 0.4**	**2.1 ± 0.6**	**0.0005**	**1.7 ± 0.5**	**2.0 ± 0.6**	**0.04**
Avg hip flexion moment (Nm/kg)	1.02 ± 0.25	0.97 ± 0.35	0.37	1.21 ± 0.39	1.19 ± 0.27	0.54
Avg knee flexion moment (Nm/kg)	**0.71 ± 0.26**	**0.97 ± 0.32**	**0.002**	**0.75 ± 0.24**	**0.97 ± 0.23**	**0.0002**
Avg ankle DF moment (Nm/kg)	**0.70 ± 0.27**	**0.86 ± 0.31**	**0.001**	**0.65 ± 0.20**	**0.74 ± 0.23**	**0.04**
Hip power absorption (J/kg)	0.80 ± 0.27	0.83 ± 0.32	0.61	1.12 ± 0.42	1.16 ± 0.38	0.56
Knee power absorption (J/kg)	**0.83 ± 0.35**	**1.36 ± 0.46**	**<0.0001**	**1.00 ± 0.38**	**1.38 ± 0.43**	**0.0001**
Ankle power absorption (J/kg)	**0.56 ± 0.19**	**0.77 ± 0.31**	**0.003**	**0.52 ± 0.20**	**0.73 ± 0.21**	**0.0003**
Peak hip internal rotation (degrees)	5.8 ± 1.9	8.0 ± 1.8	0.29	11.4 ± 5.2	10.3 ± 7.1	0.57
Peak hip adduction (degrees	-6.9 ± 6.1	-8.6 ± 6.4	0.37	-9.8 ± 5.1	-11.6 ± 5.6	0.22
Min knee varus (degrees)	-1.6 ± 6.5	-1.3 ± 5.2	0.77	0.2 ± 4.3	0.3 ± 3.8	0.91
Min knee varus moment (Nm/kg)	-0.51 ± 0.23	-0.43 ± 0.26	0.31	-0.51 ± 0.26	-0.39 ± 0.32	0.06

Table 2: Kinematic and kinetic changes between visits during the vertical drop jump.

	Change			
	Op	**P**	**Non-op**	**P**
Peak hip flexion (degrees)	**10.4 ± 15.8**	**0.007**	**10.4 ± 15.9**	**0.007**
Peak knee flexion(degrees)	**9.5 ± 13.6**	**0.005**	**7.6 ± 12.7**	**0.01**
Peak ankle dorsiflexion (degrees)	-0.7 ± 6.6	0.62	-1.0 ± 6.3	0.48
Peak vertical GRF (multiples of body weight)	0.18 ± 0.44	0.08	-0.11 ± 0.68	0.47
Avg hip flexion moment (Nm/kg)	**0.19 ± 0.29**	**0.008**	**0.22 ± 0.32**	**0.004**
Avg knee flexion moment (Nm/kg)	0.03 ± 0.16	0.33	-0.004 ± 0.27	0.95
Avg ankle DF moment (Nm/kg)	-0.04 ± 0.13	0.17	-0.12 ± 0.30	0.11
Hip power absorption (J/kg)	**0.33 ± 0.42**	**0.002**	**0.33 ± 0.51**	**0.008**
Knee power absorption (J/kg)	**0.18 ± 0.29**	**0.01**	0.02 ± 0.39	0.82
Ankle power absorption (J/kg)	-0.04 ± 0.16	0.22	-0.04 ± 0.37	0.63
Peak hip internal rotation (degrees)	**5.6 ± 8.3**	**0.006**	2.3 ± 7.8	0.19
Peak hip adduction (degrees	**-2.9 ± 6.2**	**0.04**	-3.0 ± 7.0	0.07
Min knee varus (degrees)	1.8 ± 5.3	0.15	1.6 ± 3.7	0.06
Min knee varus moment (Nm/kg)	0.002 ± 0.17	0.96	0.04 ± 0.36	0.61

DISCUSSION

During early rehabilitation pediatric and adolescent athletes with ACLR reduce flexion and loading of the knee and ankle on their operative limb, possibly representing an avoidance mechanism. Motion and loading at the knee increase over time, but remain reduced relative to the contralateral side 6-10 months post-operatively. Increased bilateral hip flexion motion, moments, and power absorption may indicate improvements in proximal control as rehabilitation progresses, which may help compensate for persistent deficiencies at the knee.

REFERENCES

1. Sanders TL et al (2016) Am J Sports Med, 44(6):1502-7.
2. Paterno MV et al (2014) Am J Sports Med, 42: 1567-73.

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DISCLOSURE STATEMENT

One author is a consultant for Arthex and Ceterix; neither company was involved in any aspect of this study. The remaining authors have no conflicts of interest to declare

Asymmetrical Intra-Limb Coordination at Return to Sport following Anterior Cruciate Ligament Reconstruction

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INTRODUCTION

Most anterior cruciate ligament (ACL) ruptures occur during dynamic movements of landing and cutting maneuvers. Currently, discrete variables (valgus knee displacement, weak hip abductor strength, etc.) are used in biomechanical models of lower extremity injuries despite the complex coordination of multiple limb segments that influence proper biomechanical profiles during functional movements. While current prevention and rehabilitation training [1,2] that target these discrete measures have been effective, there have been limited investigations on how coordination influences ACL injury risk. Previous investigations on joint coordination have primarily focused on the mechanics of gait patterns as related to injury [3] and have proposed that reductions in joint coordination variability limits an individual's ability to execute flexible and adaptable movement patterns – a potential indicator of an injured state. The current study examined interlateral asymmetry of intra-limb coordination at return to sport following ACL reconstruction.

CLINICAL SIGNIFICANCE

Traditional biomechanical measures of lower extremity risk factors fail to capture inter-segment coordination and identification of interlateral asymmetries of joint coordination may be missing a component of current prevention and rehabilitation approaches to ACL injuries.

METHODS

Eleven young female athletes with an average age of 14.6±1.07 years, height of 166.66±4.25 cm, mass of 64.45±9.42 kg, and an IKDC score of 92±6.5 volunteered for this study. Each participant had experienced unilateral ACL reconstruction and completed rehabilitation to the point of return to sport clearance. Thirty-three reflective markers were placed over the lower extremity and trunk of each participant. Participants completed 3 trials of drop (30" box) vertical jump landings on a dual force platform (AMTI, Boston, MA) setup that was surrounded by and integrated with a motion capture system (Qualysis, Sweden). Three-dimensional joint and segment angles were calculated with respect to lab coordinates. Three joint couplings (foot-shank, shank-thigh, thigh-pelvis) were created from segment angle data and used to create relative motion (angle-angle) plots as a function of anatomical (sagittal, frontal, and transverse) plane (Figure 1). Statistical calculations from relative motion (RM) plots were computed from a modified vector-coding technique [4] and included: (1) relative motion area and (2) vector-coding that calculated an angle, with respect to the right horizontal, of the vector formed between two consecutive data points on the RM plot. Statistical analyses included separate repeated-measures ANOVAs with leg, joint coupling, and anatomical plane as the repeated factors. An alpha value of 0.05 was used to define statistical significance.

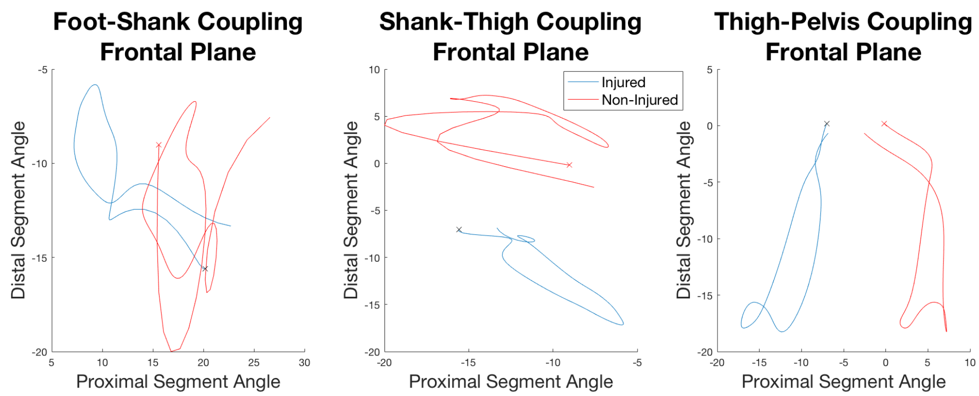

Figure 1: Representative relative motion plot from frontal plane analysis for foot-shank (left), shank-thigh (middle), and thigh-pelvis (right) joint couplings. The 'X' symbol marks initial contact from drop landing.

RESULTS

The RM area differed across joint couplings ($p<0.001$) with greater area in shank-thigh coupling than other couplings. The sagittal plane RM area was significantly greater than transverse and frontal planes ($p<0.001$). A significant triple interaction (leg * coupling * plane) revealed asymmetrical thigh-pelvis coupling in the sagittal plane with smaller area found in the injured limb ($p<0.001$). The results of vector-coding also showed a significant triple interaction (leg * coupling * plane) with differences between the limbs in shank-thigh in the transverse plane and thigh-pelvis in the sagittal plane ($p<0.001$).

DISCUSSION

At the time of return to sport, asymmetrical joint coordination between the injured and non-injured limb was identified suggesting differential movement patterns. The smaller RM area found in the injured limb may indicate restricted movement patterns that reflects a protective strategy that limits flexibility and avoids painful movements. This asymmetry needs to be explored across different functional movements and compared with existing injury risk models to better understand how ACL prevention and rehabilitation training can impact joint coordination and potential reduce injury risk. Also, comparisons to healthy control subjects are needed to address whether compensatory joint coordination exist within the non-injured limb.

REFERENCES

[1] Yoo, J.H. et al. (2010) Knee Surgery, Sports Traumatology, Arthroscopy 18 (824–830.
[2] Hewett, T.E. et al. (2016) American Journal of Sports Medicine
[3] Hamill, J. et al. (1999) Clinical Biomechanics. 14 297–308.
[4] Sparrow, W.A. et al. (1987) Journal of Motor Behavior 19 115–29.

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LANDING MECHANICS IN UNINJURED ADOLESCENT ATHLETES: A REVIEW OF SYMMETRY IN SELECTED KNEE KINEMATIC AND KINETICS

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INTRODUCTION: ACL reconstruction (ACLR) is a common orthopaedic surgery and requires a lengthy rehabilitation program if a patient wishes to successfully return to sports (RTS) at the pre injury level of sports participation [1]. The formal assessment of an athlete's readiness to RTS is a critical point in the patient's progress. Currently, one of benchmarks used at this institution for an adolescent athlete to receive a physician's clearance to RTS is attaining correct and symmetrical landing mechanics in a variety of functional tasks. However, symmetry in these tasks has not been documented in the healthy, uninjured adolescent athlete population. Therefore, the purpose of this study was to evaluate the symmetry in landing mechanics in uninjured adolescent athletes while performing functional assessment tasks using motion analysis techniques.

CLINICAL SIGNIFICANCE: It is critical to understand the symmetry in landing mechanics in uninjured, age matched athletes in order to appropriately interpret symmetry in RTS assessment data in an ACLR patient.

METHODS: The control data collection for this study was part of a larger study of ACLR in adolescent athletes. All athletes completed functional tasks which included: the drop vertical jump (DVJ), single leg long hop (LH), triple hop (TH), cross-over triple hop (CH) and running (Run). Motion data was collected using a 12 camera Vicon MX system (VICON, Los Angeles, CA) while ground reaction forces were collected using 5 force plates (AMTI, Watertown, MA) following standard procedures [2]. A single representative trial was selected for analysis of the DVJ and running tasks, while a successful trial with the longest jump distance from each side was selected for the LH, TH and CH.

Descriptive statistics and the limb symmetry index (LSI=[(non-preferred ÷ preferred)*100)]) were computed to compare the preferred to the non-preferred kicking leg for peak knee flexion (PKF), peak knee extensor moment (PKEM) and peak knee power absorption (PKPA) during the knee loading phase of each task. Paired t-tests (α=0.05) were used to determine if there were statistically significant differences between sides. A LSI between 90% and 110% was considered symmetrical for the selected parameters.

RESULTS: Thirty control subjects (14 female, 15±1 years, height 169±10 cm, weight 59±14 kg) completed the protocol. The means of the peak kinetic measures tended to be greater on the preferred side, however statistically significant differences were found only for the LH (Table 1). LSI calculations showed that asymmetry was present in the majority of control subjects for PKF in the LH and TH, and for PKEM and PKPA in all functional tasks (Table 1). For control subjects with asymmetry, the preferred kicking side was not always greater than the non-preferred kicking side (i.e. LSI ≤ 90%). The range of the LSIs for control subjects was greater in PKEM and PKPA than in PKF for all functional tasks (Fig. 1).

DISCUSSION: Symmetry in landing mechanics is a primary criterion in the current standard of care in RTS decisions post ACLR. The results of this study demonstrated no side to side differences in mean peak sagittal knee kinematic and kinetic variables with the exception of the LH. Interestingly, this is the only functional task that analyzes a final landing instead of a landing immediately followed by a takeoff. However, when evaluating individual control subjects, asymmetry of >10% was present in many for PKF in the LH, TH and CH. Also, the majority of control subjects had asymmetrical knee kinetics for all functional tasks.

The extent of asymmetry varied across parameters and functional tasks, with the greatest symmetry measured in PKF of the DVJ. This suggests that some functional tasks may be more informative than others for determining the presence of asymmetry. Further investigation into the utility of symmetry in peak knee kinematics and kinetics as the optimal parameters for evaluating ACLR patients is required, as perhaps a threshold limit would be more useful in making RTS decisions. Monitoring this cohort for future injury would determine if those subjects with an asymmetry between sides are at an increased risk of an ACL tear.

Table 1. Mean ± SD knee kinematics and kinetics and incidence of LSI findings.

	PKF (°)				PKEM (Nm/Kg)				PKPA (W/kg)			
	Pref Knee	Non-Pref Knee	% of subjects LSI ≤90%	% of subjects LSI ≥110%	Pref Knee	Non-Pref Knee	% of subjects LSI ≤90%	% of subjects LSI ≥110%	Pref Knee	Non-Pref Knee	% of subjects LSI ≤90%	% of subjects LSI ≥110%
DVJ	87 ± 15	87 ± 15	0%	3%	1.57 ± 0.35	1.45 ± 0.40	40%	17%	11.35 ± 3.72	11.26 ± 4.56	37%	43%
LH	58 ± 8	55 ± 10	37%	20%	1.80 ± 0.59	1.57 ± 0.59	53%	27%	13.46 ± 5.90	10.9 ± 5.47	57%	30%
TH	56 ± 6	53 ± 7	33%	23%	1.92 ± 0.46	1.80 ± 0.51	43%	20%	12.30 ± 3.39	10.75 ± 4.27	53%	20%
CH	57 ± 7	57 ± 7	17%	17%	1.90 ± 0.58	1.75 ± 0.43	37%	23%	10.13 ± 4.17	8.87 ± 4.19	57%	30%
Run	44 ± 5	43 ± 5	23%	7%	1.64 ± 0.52	1.64 ± 0.55	37%	50%	9.66 ± 4.41	10.29 ± 4.90	30%	47%

PKF = Peak knee flexion; PKEM = Peak knee extensor moment; PKPA = Peak knee power absorption; Non-Pref = non preferred kicking side; DVJ = drop vertical jump, LH = single leg long hop; TH = single leg triple hop; CH = single leg crossover triple hop.
Grey shaded cell = $p \leq 0.05$ paired t-test between sides. A LSI <100% indicates the preferred side is greater than the non-preferred side

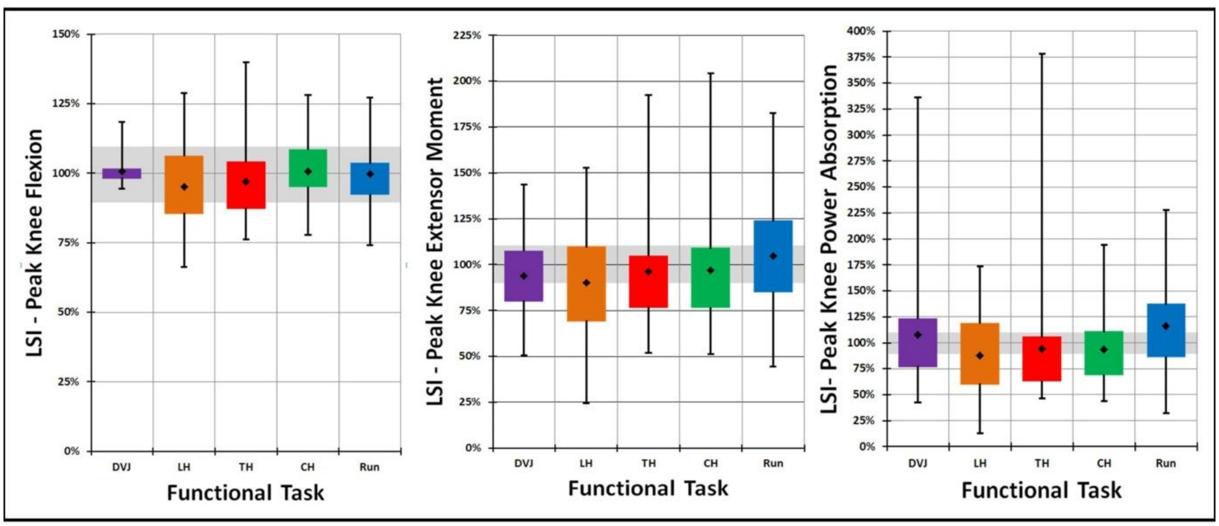

Figure 1: Box and whisker plots of the LSI distribution of each knee parameter for the functional tasks. ♦=Mean, Grey Band = 90% ≤ LSI ≤ 110%

REFERENCES:

1. McCullough, K.A. et al. (2012) Am J Sports Med, 40(11): 2523-2529.
2. Davis, R. et al. (2000) Analysis Biomedical Engineering Handbook, pp. 1-9.

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COMPARISON OF BALANCE CONTROL DURING DUAL-TASK WALKING IN ADOLESCENT ATHLETES AFTER CONCUSSION: A MULTICENTER STUDY

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INTRODUCTION

The decision of when an athlete is ready to return to play (RTP) after concussion remains among the most difficult in sports medicine. Currently, RTP decisions are based on resolution of clinical symptoms and basic cognitive, neuropsychological, and standing balance tests. However, assessments of higher level activities such as walking in combination with simultaneous cognitive tasks have shown lingering impairments despite normalization of basic functions [1]. Such dual-task assessments are not typically incorporated into RTP decisions, but may be useful for revealing remaining balance control deficits. Therefore, the purpose of this study was to determine if concussed adolescent athletes exhibit remaining deficits in dual-task balance control at the time of RTP clearance. Teenage and younger athletes account for two-thirds of sports related concussions, but concussion recovery has not been studied in this younger age group. We hypothesized that adolescent athletes cleared to RTP based on current clinical criteria would walk significantly slower with greater medial-lateral (ML) sway during cognitive function tasks compared to non-concussed peers.

CLINICAL SIGNIFICANCE

In contrast to adult and older adolescent athletes [1], this group of adolescent athletes demonstrated no difference in dual-task balance control compared to controls at the time of RTP. This may indicate that they were cleared to RTP appropriately under the more conservative approach to concussion management in this age group which involves a longer rest period and later RTP. Alternatively, given the variability among subjects in the measures studied, these results may reflect a need to develop more sensitive quantitative outcome measures utilizing motion analysis.

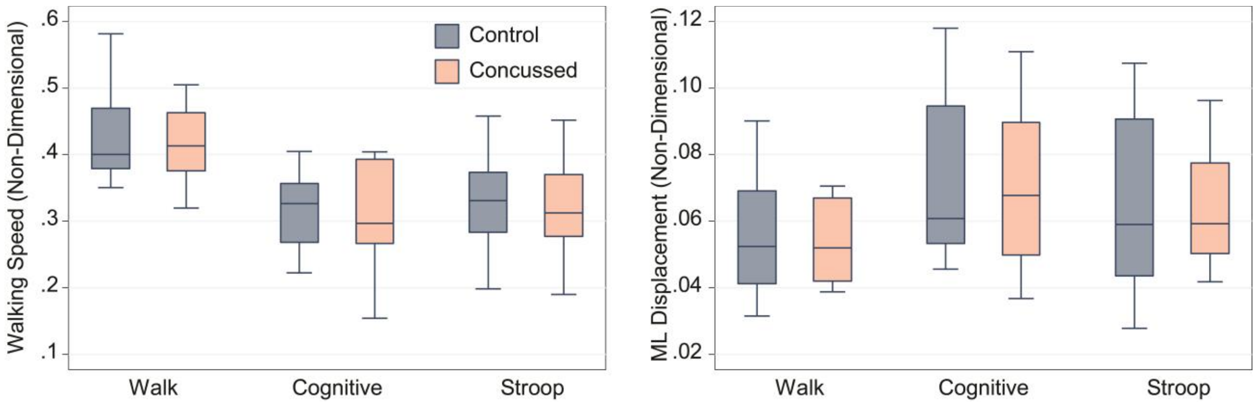
METHODS

This prospective multicenter study examined 16 concussed patients (9 male) and 15 controls (9 male) ages 11-17 years (mean 14.2, SD 1.9) after written informed consent was obtained. Patients were examined within 1 week of RTP clearance. All subjects participated in a single test session in which they underwent motion analysis testing while walking at a self-selected speed during 3 tasks: 1) walking only, 2) a simple cognitive task reciting the months of the year backwards and 3) a continuous 1-second interval audio Stroop test. Walking velocity and ML center of mass displacement and velocity adjusted for body size using non-dimensional normalization were compared among groups and tasks using two-way ANOVA (including interaction term) with pairwise Bonferroni adjusted posthoc tests.

RESULTS

The average time from concussion to testing was 57 days (SD 46, range 12-170). Walking speed and ML displacement differed significantly among tasks ($p<0.01$) but not between groups ($p>0.55$) (Table & Figure). Walking speed was slower for both dual-task conditions compared with walking only ($p<0.05$). ML displacement was larger for the cognitive dual-task compared with walking only ($p<0.05$). ML velocity did not differ significantly with task or group ($p>0.14$). There was substantial between and within-subject variability in walking speed and COM motion in both groups.

Figure & Table: Non-dimensional walking speed and COM motion by task and group



	Control	Concussed	P
Walking Only			
Walking Speed	0.42 ± 0.06 (0.35, 0.58)	0.41 ± 0.06 (0.32, 0.51)	0.62
ML Displacement	0.054 ± 0.016 (0.30, 0.59)	0.054 ± 0.012 (0.32, 0.51)	0.98
ML Velocity	0.024 ± 0.007 (0.01, 0.04)	0.027 ± 0.007 (0.02, 0.04)	0.41
Cognitive			
Walking Speed	0.31 ± 0.05 (0.22, 0.40)	0.31 ± 0.08 (0.15, 0.40)	0.98
ML Displacement	0.072 ± 0.026 (0.20, 0.41)	0.070 ± 0.023 (0.16, 0.41)	0.82
ML Velocity	0.029 ± 0.008 (0.02, 0.04)	0.030 ± 0.008 (0.02, 0.04)	0.52
Audio Stroop			
Walking Speed	0.33 ± 0.07 (0.20, 0.46)	0.32 ± 0.07 (0.32, 0.51)	0.61
ML Displacement	0.062 ± 0.025 (0.20, 0.46)	0.063 ± 0.017 (0.19, 0.45)	0.89
ML Velocity	0.027 ± 0.008 (0.02, 0.04)	0.029 ± 0.008 (0.02, 0.04)	0.29

Values are presented as Mean ± SD (range). The P-values shown are from the post-hoc tests.

DISCUSSION

In contrast to older athletes, this cohort of adolescent athletes demonstrated no difference in dual-task balance control compared to controls at the time of clearance to RTP. This may reflect the more conservative management, and thus the longer time to RTP, for this younger age group who returned an average of 2 months after concussion compared with 2 weeks for older athletes. Because of the large variability in walking speed and COM motion among subjects, including within the control group, it may be difficult to use these measures to assess RTP readiness for individual patients. More sensitive measures are needed to determine more conclusively whether adolescent athletes are indeed being treated optimally for concussion.

REFERENCES: [1] Howell et al., Arch Phys Med Rehab 94, 1513-1520, 2013.

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THE EFFECT OF KNEE SUPPORT ON THE KNEE LOADS AND MOTIONS OF BASEBALL CATCHERS IN DEEP SQUATS

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INTRODUCTION

It is estimated that 11.5 million baseball players will take the field this spring in the United States [1]. While not a contact sport, injury prevention and safety are still major areas of concern for amateur baseball players. One example of this is the risk to catchers' knees in both baseball and softball, due to the repeated and prolonged squatting required to play this position, especially for players who begin in little league and progress over the years to college and beyond. It has been determined from a cadaver model that deep knee flexion results in forces across the knee that are 80% higher than normal [2]. This is enough force to reach the damage limits of articular cartilage. "KneeSavers" (Trademark by Easton. CA) have been developed to try and alleviate some of the knee loads. These are foam wedges worn on the dorsal side of the calf that help support the catchers' weight as they squat. This work develops a method to quantify the decreased loads associated with use of KneeSavers.

CLINICAL SIGNIFICANCE

The models developed here will simulate knee loads with and without KneeSavers, giving physicians more information to predict deterioration of the knee joint. With this information a determination could be made about the length of playing time catchers should participate in, in order to keep them performing at their best. Determining the efficacy of a KneeSaver type device could lead to lower injury rates among catchers and better knee health throughout their life.

METHODS

Experimental data was collected for ten subjects, two females and eight males, using Vicon and the plugin gait marker set. Subjects were asked to squat into a catcher's stance in three conditions, with no device and while wearing the instrumented wedges (Fig. 1). Forces carried by the knee support were measured using custom developed load cells with marker clusters attached to record the position and direction of the loads applied during the squat. Under each condition the subjects were asked to step onto the force plates, squat and hold to the count of two, rise, and step backwards off of the plates. The force data and maker location data was then exported for modeling in MSC.Adams with the LifeModeler plug in.

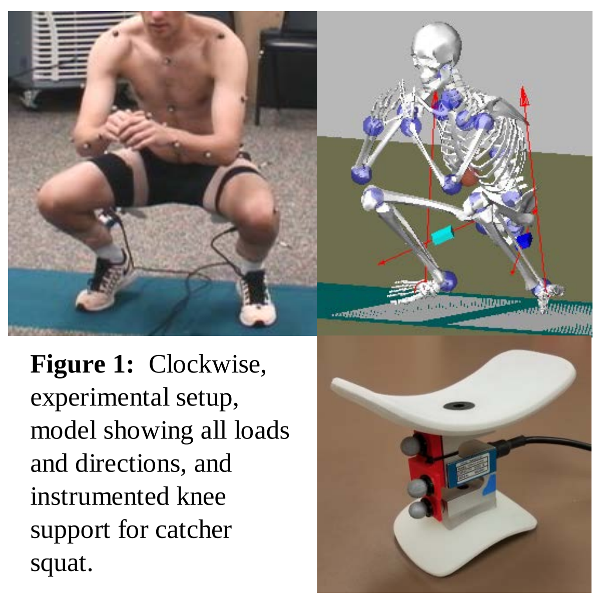

Figure 1: Clockwise, experimental setup, model showing all loads and directions, and instrumented knee support for catcher squat.

The models were run to calculate knee torques required for three squats conditions: 1) no knee support, 2) support present but measured forces not used, and 3) support present and forces used. Joint torques and angles from these models were then exported for analysis in Matlab.

RESULTS

From analysis of the models, there is a significant reduction in torque while wearing knee support during deep squats. Specifically, a 47% reduction was seen on the dominant side, and a 64% reduction on the non-dominant side (Fig. 2). This was determined by comparing the torque on the knee joint in the model with the subject wearing the load cell knee savers without the force of the load cells applied versus applying the load cell force to the body. We noted a statistically significant difference in knee angle between the subjects wearing the load cells and the subjects wearing nothing. The difference in the means between these conditions was 6.4 degrees.

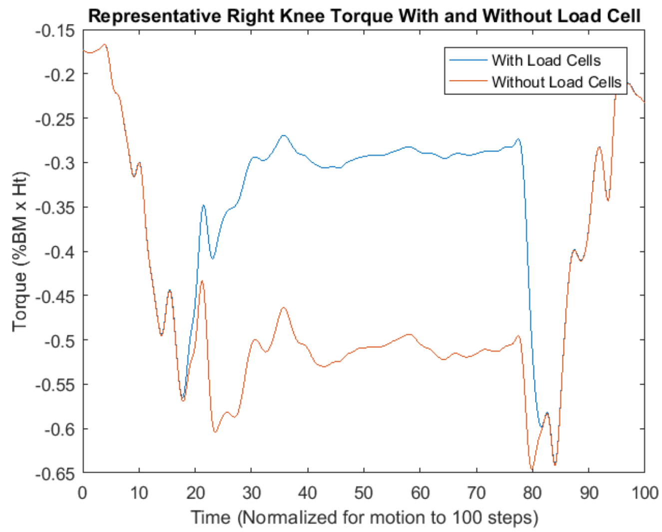

Figure 2: Typical results of knee torque during a deep squat, red with no support, blue with support.

DISCUSSION

The output torque from the no support condition was validated against previously reported knee torque during a squat at 90° [3], results of both models were similar. The reduction in torque at the knee joint is a critical result. This decrease in force across the knee decreases the likelihood that the deep knee flexion necessary to squatting in the catching position will reach the damage limit for the cartilage in the knees. The reduction in knee torque required to hold a deep squat position with support will result in a much greater reduction in the compressive force at the knee due to the small moment arm of the quadriceps about the knee. This compressive force often results in degradation/tears of the meniscus. This ultimately would save the catcher's knees from injury and extend the catcher's ability to play baseball.

Regarding the difference in knee angles between subjects wearing nothing and wearing the knee savers, while this angle was statistically significantly different this difference is not physically different. This means that this difference would not be obvious to the player, as it is too small for them to consistently be able to adjust that amount. It was then concluded that the reduction in torque between the conditions examined would be similar to the reduction in torque between wearing savers and wearing nothing. How the KneeSavers effect reaction time becomes a potential reason for not wearing them at higher levels, and should be investigated in the future.

REFERENCES

1. Futterman, M. (2011) Wall Street Journal.
2. Thambyah, A. et al. (2005) ME&P, 27: 329 – 335.
3. Han, S. et al. (2013) JHK, 39: 59 – 66.

DISCLOSURE STATEMENT

We have no disclosures or conflicts of interest.

Podium Session #3

FOOT/ANKLE

MODERATED BY: **Roy Davis, PhD:** Director, Motion Analysis Center
Shriners Hospital for Children, Greenville, SC

Howard Hillstrom, PhD: Director, Leon Root, MD Motion Analysis Laboratory
Hospital for Special Surgery, New York, NY

1. **Plantar Pressures Following Surgical Intervention for Clubfoot: Intermediate Follow Up at 5 Years of Age**
   Ashley Erdman, Kelly Jeans, Lori Karol

2. **Outcomes Following Treatment for Idiopathic Clubfoot at Age 10yrs: Gross Motor Function, Strength & PODCI**
   Kelly Jeans, Lori Karol, Karina Zapata, Ashley Erdman

3. **Segmental Foot and Ankle Kinematics of Children with Charcot-Marie-Tooth Disease**
   Karen Kruger, Joseph Krzak, Adam Graf, Maria Romano, Haluk Altiok, Gerald Harris

4. **Foot Deformities and Gait Deviations in Children with Arthrogryposis**
   Lucio Perotti, Chris Church; Celina Santiago; Nancy Lennon; John Henley; Maureen Donohoe, Kathryn Fazio, Freeman Miller, Louise Reid Nichols

5. **Partitioning Ground Reaction Forces for Multi-Segment Foot Kinetics**
   Dustin Bruening, Kota Takahashi

6. **A Pedobarograph Measure to Assess Center of Pressure**
   Bruce MacWilliams

7. **Comparison of Three-dimensional Multi-Segment Foot Models**
   Chris Church, Colton Takata, Kristen Nicholson, Tim Niiler, Brian Chen, Nancy Lennon, Julie Sees, Freeman Miller, John Henley

PLANTAR PRESSURES FOLLOWING SURGICAL INTERVENTION FOR CLUBFOOT: INTERMEDIATE FOLLOW UP AT 5 YEARS OF AGE

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INTRODUCTION

Nonoperative treatment methods for clubfoot (CF) have reduced the rate of surgical intervention while providing fewer kinematic abnormalities compared to surgically corrected feet. Previous work shows clubfeet that require surgical correction exhibit residual deformity, stiffness and pain and are four times as likely to develop calcaneus gait as a result of over-lengthening and extensive release [1-3]. The purpose of the current study was to assess plantar loading in clubfeet initially treated nonoperatively that required surgical correction by age 5 years and to compare their outcomes to age-matched controls.

CLINICAL SIGNIFICANCE

Longitudinal outcomes following surgical management of CF will help clinicians better understand the effects on patient function and potentially guide future treatment decisions.

METHODS

Plantar pressure data were collected at age 5 years in patients with CF initially treated nonoperatively who ultimately required surgical intervention before their 5 year visit. Surgical feet were subdivided into groups based on the extent of surgical release: Limited surgery (posterior release, anterior tibial tendon transfer, lateral column shortening) and PMR (posteromedial release ± additional procedures). A minimum of 5 trials were collected for each affected foot using the *Emed System* (Novel, Munich, Germany). A representative trial was selected for analysis and the *Novel PRC Automask* was applied to divide the foot into the following regions: medial/lateral hindfoot (HF), medial/lateral midfoot (MF), 1st metatarsal head, 2nd metatarsal head and 3-5th metatarsal heads. When necessary, manual corrections were made to the automask using a static plantar picture for reference [3]. The following variables were analyzed: Peak Pressure (**PP**, N/cm^2), Maximum Force (**MF%**, % body weight), Contact Area% (**CA%,** % of total foot), and Contact Time% (**CT%**, % of roll over process). The hindfoot-forefoot angle (**HFA**) and the displacement of the center of pressure (**COP**) line were also calculated [3]. The limited surgery and PMR groups were compared to 20 age-matched controls using an ANOVA with a post-hoc Tukey test (α=0.05).

RESULTS

Plantar pressure data from 122 surgical CF (58 PMR, 64 limited) were included. Reduced MF% in the lateral HF of the PMR group was the only significant difference when compared to the limited surgery group (*Table 1*). Compared to controls, both surgical groups showed significant differences across variables. PP was reduced in the medial and lateral HF while MF% was reduced in the medial HF, which documents decreased HF loading due to restricted dorsiflexion. All variables were increased in the lateral MF compared to controls and CT% was increased in the medial MF consistent with a degree of residual varus. In the 1st metatarsal region, all variables were significantly reduced in both surgical groups, whereas CA%, CT% and MF% were all increased in the 3rd - 5th metatarsal region. Significant differences seen only in the PMR group were increased CT% in both the medial and lateral HF consistent with poor heel rise at push-off. CA% was also reduced in the 2nd metatarsal in the PMR group indicating increased lateralization compared to controls. The HFA showed that the FF is significantly medial relative to the HF in both groups compared to controls (limited surgery 167.9° ± 7.8°, PMR 165.1° ± 7.5° and controls 174.2° ± 6.3°; P < 0.001). The COP line moved significantly more lateral in both surgical groups compared to controls (limited surgery 10.3 ± 7.4, PMR 11.8 ± 6.2, and controls -2.6 ± 3.1; P < 0.001).

DISCUSSION

Plantar pressures showed few differences between feet that require limited surgery and those that underwent a PMR at age 5 years. Feet requiring a PMR frequently demonstrate weakness or insufficiency of the gastrocsoleus as was demonstrated by increased CT% in the HF which correlates with the increased likelihood of calcaneus in surgically released CF [1]. Lateralization is still evident in both surgical groups shown by a shift in the COP line, as well as an increase in HFA and reduced pressure in the 1st metatarsal region. Similar to the nonoperative cohort at age 5, compromise of the medial arch was observed in the surgical groups identified by increased CT% in the medial MF with increasing likelihood as the level of surgery becomes more involved [4]. Although surgery improves the position of a clubfoot, the inherent muscle weakness as well as any postoperative stiffness prevents normalization of plantar pressures, despite a good clinical result. Surgeons should continue to strive to correct and maintain correction in clubfeet nonoperatively, as those feet that undergo intra-articular surgery may improve, but do not normalize following surgery when assessed at the age of 5.

TABLE 2. Plantar Pressures Comparing Limited surgery (n=64), PMR (n=58) and Control (n=40)

	Medial HF		Lateral HF		Medial MF		Lateral MF		1st Met.		2nd Met.		3-5th Mets.	
	Mean	±SD	Mean	±SD	Mean	±SD	Mean	±SD	Mean	±SD	Mean	±SD	Mean	±SD
PP (N/cm^2)														
Limted	149.5	81.9	138.0	61.9	82.1	32.4	111.0	35.3	79.1	51.1	123.2	48.0	189.8	105.5
Full	165.1	74.1	144.7	47.3	79.8	28.6	122.0	53.9	61.6	31.6	111.8	37.4	187.5	91.4
Control	*271.3*	*111.4*	*226.4*	*92.3*	*67.5*	*22.9*	*67.8*	*18.4*	*111.3*	*38.8*	*158.5*	*34.2*	*156.3*	*37.4*
P	<0.001†,*		<0.001†,*		0.0353†		<0.001†,*		0.001†, <0.001*		0.001†, <0.001*		0.150	
MF% (%BW)														
Limted	30.5	67.3	33.4	9.3	6.4	6.6	39.6	15.1	10.5	8.7	16.4	6.8	49.3	15.1
Full	31.2	12.0	28.8	9.7	7.3	6.7	38.8	15.1	11.1	8.6	15.8	6.4	51.9	17.4
Control	*47.5*	*11.8*	*35.5*	*7.8*	*4.7*	*5.8*	*14.7*	*8.2*	*21.1*	*7.6*	*23.4*	*4.5*	*38.1*	*9.2*
P	<0.001†,*		0.001*, 0.017^		0.160		<0.001†,*		<0.001†,*		<0.001†,*		0.001†, <0.001*	
CA%														
Limted	11.3	2.5	11.7	2.7	4.8	3.8	21.0	3.8	7.7	3.7	8.7	2.0	21.9	6.7
Full	11.8	2.0	11.9	2.1	4.8	3.6	21.0	4.0	6.7	2.9	8.3	1.8	22.3	5.5
Control	*12.2*	*1.1*	*12.4*	*1.0*	*3.6*	*2.7*	*14.7*	*4.4*	*10.8*	*1.8*	*9.4*	*1.3*	*18.5*	*2.6*
P	0.110		0.308		0.165		<0.001†,*		<0.001†,*		0.010*		0.008†, 0.003*	
CT%														
Limted	52.6	16.7	54.5	16.5	52.8	18.0	75.2	8.3	68.3	26.3	81.5	17.2	90.4	5.8
Full	57.8	18.2	59.2	17.4	56.3	19.8	77.4	8.3	60.6	24.8	78.0	16.5	90.4	5.3
Control	*49.1*	*9.0*	*48.3*	*8.7*	*39.1*	*9.9*	*55.4*	*13.4*	*80.2*	*9.4*	*82.6*	*7.4*	*84.9*	*5.2*
P	0.022*		0.002*		0.001†, <0.001*		<0.001†,*		0.028†		0.265		<0.001†,*	

† Limited surgery different from Control; *PMR different from Control; ^PMR different from Limited surgery

REFERENCES

1] Karol, LA JPO 2005. 2] Noonan, KJ JAAOS 2003. 3] Jeans, KA JPO 2010. 4] Jeans, KA JPO 2016.

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OUTCOMES FOLLOWING TREATMENT FOR IDIOPATHIC CLUBFOOT AT AGE 10YRS: GROSS MOTOR FUNCTION, STRENGTH & PODCI

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INTRODUCTION

Over the last 15 years, the evidence supporting a nonoperative (NO) approach in the treatment of idiopathic clubfoot (CF) has grown. At age 5 yrs, Karol et al. reported a positive correlation of ankle power and gross motor function. [Karol 16] As ankle power decreased, so did performance on the Peabody developmental motor scales including balance and locomotion tasks (running, jumping and hopping). The purpose of this study is to ascertain if gross motor function is different at age 10yrs between CF patients that remain NO and those that underwent surgical correction and to determine if a surgical approach is a significant factor in functional outcomes. Is there a relationship between functional outcome measures and parent perception?

CLINICAL SIGNIFICANCE

By treating clubfoot with a NO approach, clinicians try to correct the foot's position with the goal of reducing the incidence of surgical correction needed to achieve a plantigrade, pain-free and functional foot. Objective measures are needed to fully appreciate how the foot functions when balancing, walking, running, hopping, and jumping in the developing child.

METHODS

As part of a prospective IRB approved study, at 10yr of age, children with idiopathic clubfoot were invited to complete a gross motor function assessment (BOT-2) and isokinetic strength testing of plantarflexion (PF) while the accompanying parent completed the Pediatric Outcomes Data Collection Instrument (PODCI). All patients were initially treated NO with either the Ponseti method or the French physiotherapy technique. Some feet required surgical (SX) correction due to residual deformity or recurrence. A secondary analysis was performed comparing feet treated with *PMR*, limited posterior release (*PR*) and those that only required tendon release, transfer and/or lengthenings (*SOFT*).

The Bruininks-Oseretsky Test of Motor Proficiency 2nd (BOT-2) was administered by a trained physical therapist and scored according to age-based norms. The following subscales were assessed: bilateral coordination, balance, running speed/agility and strength. The composite motor domains include: body coordination and strength and agility. Isokinetic strength was obtained in the prone position with the foot secured to a foot plate while the participant moved maximally through PF and dorsiflexion at 60°/s. Five trials were collected and the maximum effort was used for analysis. Results from the Sport/Physical Functioning Scale (Sport/Phys) and Global Functioning Scale (GFS) of the PODCI were used to assess parent-perceived function. Since the BOT-2 and the PODCI are not "side specific" in their outcome measures, PF strength and initial Dimeglio scores of a single limb was randomly selected for the bilateral patients. Between-group comparisons were conducted using the Mann-Whitney test and correlations between BOT-2, PF strength and PODCI variables were made with Spearman's correlation coefficient. Alpha was set to 0.05.

RESULTS

There were 150 CF patients seen for BOT-2 and Biodex testing at age 10.2 yrs (range 9.7-11.5 yrs) with 90 remaining NO at the time of testing. Of the 60 SX patients, 27 underwent a *PMR*, 18 a *PR* and 15 with extra-articular surgery (*SOFT*). Initial Dimeglio scores were significantly higher in the SX (14.3±1.9) than the NO group (11.8±2.7; p<0.001). No differences were found in BOT-2 scores between NO and SX (P>0.05). Distribution of the results based on descriptive categorization (well above average, above average, average, etc) was not different between NO and SX (P>0.05; *Figure* 1) Peak PF strength was greater in the NO group (24.8±9.3) than the SX group (19.1±6.9; p<0.001). A closer look at the SX groups shows that PF strength was significantly less in the *PMR* group (16.7±6.4) compared to the *SOFT* group (22.4±7.7; P=0.023). Compared to the NO group (24.8±9.3), both the *PR* (19.9+5.7) and *PMR* (16.7±6.4) groups had significantly reduced PF strength (P<0.05). PODCI surveys were collected from 119 CF parents. Results show that SX parents (n=43) reported higher Sport/Phys (95.2±6.7) than the NO parents (n=76; 90.7±11.8; P=0.031). Peak PF was found to significantly correlate to the Balance, Strength and to both composite standard scores, Body Coordination and Strength/Agility (P<0.05). *Table 1* No differences were found in any parameter between bilateral and unilateral CF patients.

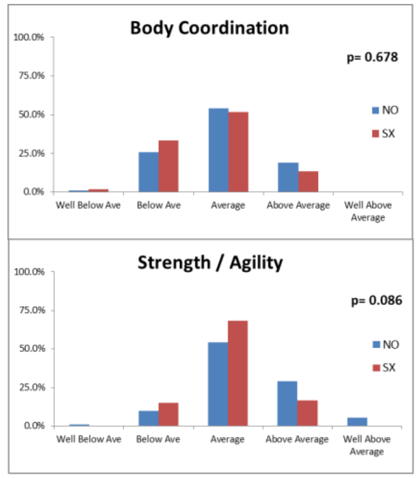

Figure 1. BOT-2 Descriptive category distribution.

	Balance	Strength	Body Coordination	Strength / Agility
Peak PF (r)	0.190	0.175	0.216	0.182
P	0.020	0.033	0.008	0.026

Table 1. Spearman's Correlation coefficient results

DISCUSSION

Few differences were found between patients that had surgery and those who remained NO at 10 years of age. Plantarflexion strength was weaker in the surgical patients, specifically the feet that undergo inter-articular release (PMR, PR). However functionally, as a group, a score of "average" or higher was achieved in 70% of CF patients for Body Coordination and 87% of patients for Strength/Agility on the BOT-2 at age 10yrs. Parents of SX patients scored their child's ability significantly higher for Sport/Phys than parents of NO children; however, both groups scored their children 90% or above, which indicates a good outcome. Correlations show a weak but significant relationship between PF strength and BOT-2 performance. Although PF strength shows a deficit, BOT-2 functional assessments indicate that 70-87% of patients are average or above compared to their peers. Even though PF strength measures show deficit in the surgical group, gross motor function measured with the BOT-2 show no difference between NO and SX outcomes.

REFERENCES 1] Karol LA, JPO 2016

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SEGMENTAL FOOT AND ANKLE KINEMATICS OF CHILDREN WITH CHARCOT-MARIE-TOOTH DISEASE

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INTRODUCTION

Charcot-Marie-Tooth disease (CMT) is a genetically heterogeneous disorder characterized by degeneration of the longest motor and sensory nerve fibers and commonly impacts the foot. Previous work identified foot drop, hindfoot inversion, and reduced ankle flexion moments [1-4] however, to date, there have been no three-dimensional analyses using a segmental foot model. The goal of this work is to identify segmental foot deviations during gait among patients with CMT.

CLINICAL SIGNIFICANCE

A better understanding of segmental foot dysfunction during gait in children and youth with CMT will assist in prescribing appropriate treatment and understanding gait prognosis.

METHODS

This was a retrospective study using an institutionally approved IRB protocol. Gait analysis was performed using the Milwaukee Foot Model (MFM) [5] to characterize the segmental kinematics of the foot and ankle as part of assessment for orthopaedic surgery planning. Each foot analyzed was instrumented with twelve reflective markers on bony anatomical landmarks during gait analysis. Weightbearing radiographs were taken from anterior/posterior, lateral, and modified coronal plane views and processed using custom software which aligns the marker-based reference system with the radiographic bone-based (skeletal) system.

The study population (CMT group) included 10 children (age=13.0±3.3 yrs) identified with CMT as a diagnosis and a Control Group consisting of 16 typically developing children (32 feet total, age=11.3±2.0 yrs). Statistical comparisons between the CMT and control populations were made among average angles (peak maximum and minimum values) in stance and swing and overall joint excursions. Statistical analysis was performed using a student's t-test (α=0.05) to compare differences in kinematic data between the two samples.

RESULTS

In the CMT group, the tibia was more anteriorly tilted than the Control group (Figure 1). Hindfoot deviations showed the CMT group to be more inverted and internally rotated. The forefoot was more plantarflexed and internally rotated. At the hallux, increased dorsiflexion, supination (stance only), and valgus were observed. Decreased range of motion was observed in the sagittal planes of the tibia and hallux and transverse plane of the forefoot (Table 1). There was increased range of motion in all other segments and planes with the exception of the coronal plane of the tibia, which showed no difference.

DISCUSSION

This study demonstrates the application of a skeletal-based approach to segmental foot analysis for individuals with Charcot-Marie-Tooth disease. The MFM was able to distinguish kinematic deviations in all three planes of motion. Increased plantarflexion of the forefoot and inversion of the hindfoot agree with previous work showing foot drop [1] and hindfoot inversion [2]. The variability observed in the CMT Group justifies the use of subgroups in this population [1,3,4]. This work offers insight into the underlying bony orientation and resulting kinematic deviations in a population of children with CMT. Insight provided by this data may help increase our quantitative understanding of the ambulatory kinematics of the condition as an aid to surgical planning and post-treatment follow up.

Table 1: Range of Motion (Degrees)

Segment	Plane of Motion	Control μ (S.D)	CMT μ (S.D)
Tibia relative to Global	Sagittal	70.7 (11.3)	60.6 (5.3)*
	Coronal	18.1 (14.5)	20.5 (7.9)
	Transverse	13.2 (5.4)	18.3 (5.9)*
Hindfoot relative to Tibia	Sagittal	16.9 (6.2)	21.6 (7.7)*
	Coronal	6.2 (3.1)	11 (4.8)*
	Transverse	4.9 (1.9)	7.4 (2.4)*
Forefoot relative to Hindfoot	Sagittal	13.7 (2.7)	19.1 (7.0)*
	Coronal	8.9 (2.9)	19.5 (6.5)*
	Transverse	12.4 (3.6)	9.6 (2.6)*
Hallux relative to Forefoot	Sagittal	34.3 (7.0)	21.4 (10.6)*
	Coronal	13.9 (3.7)	18.3 (7.9)*
	Transverse	12.2 (2.7)	16.1 (6.9)*

*=Statistically significant difference from Control Group

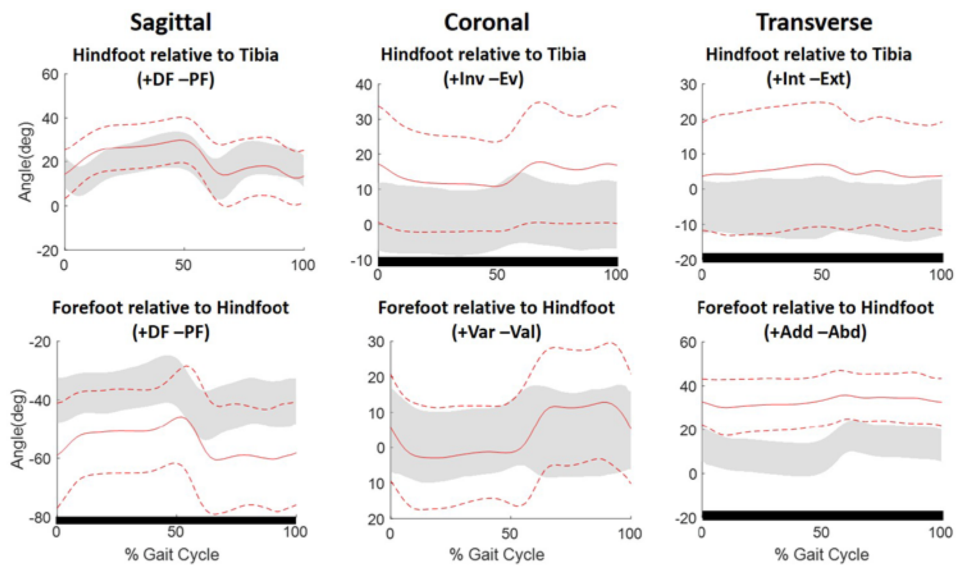

Figure 1: Selected average segmental kinematics throughout the gait cycle of CMT group (solid and dashed lines) and control group (gray band). Black bars on x axis indicate statistically significance difference in average position.

REFERENCES

[1] Don R et al. (2007) Clinical Biomechanics, 22(8):905–916. [2] Vinci P, Perelli SL. (2002) Arch Phy Med Rehabil, 83(4):513-6. [3] Ounpuu S. et al. (2013) Gait & Posture, 38:900-906. [4] Beckmann et al. (2015) Journal of Foot and Ankle Research, 8:65 [5] Long, J. et al. J. Exp. Clin. Med 2011;3(5), 239–244.

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DISCLOSURE STATEMENT

We have no conflicts of interest to disclose.

Foot Deformities and Gait Deviations in Children with Arthrogryposis

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INTRODUCTION

Arthrogryposis multiplex congenita is a congenital condition characterized by joint contractures with resulting foot deformities and gait deviations. Decreased fetal movements (fetal akinesia) are the primary cause of arthrogryposis [1]. Arthrogryposis can vary greatly with respect to limb involvement, but involvement of all four limbs is most commonly seen [2]. The most common clinically diagnosed foot deformity in arthrogryposis is clubfoot, although equinocavus foot and congenital vertical talus are also common [3]. Gait deviations in arthrogryposis have been studied and assessed clinically and improvement is often achieved with orthotics and surgery [4]. The aim of the present research study was to quantify gait deviations and foot deformities in children with arthrogryposis with detailed gait analysis including multi-segment foot kinematics and foot pressure analysis.

CLINICAL SIGNIFICANCE

Previous literature studies have rarely looked at dynamic data from gait and foot pressure analyses as well as physical exam to quantify foot deformities and gait deviations in children with arthrogryposis. Quantification of contractures and deformities may help assess outcomes and guide treatment.

METHODS

Children with arthrogryposis were evaluated retrospectively and compared to data for typically developing children. Comprehensive evaluation data included in this study consists of full gait analysis with multi-segment and single-segment foot kinematics, kinetics, foot pressure, and physical examination. In addition to evaluating the sample as a whole, children were grouped by age, orthotic use, and history of surgical intervention.

RESULTS

Forty-two children with arthrogryposis age 10 ± 5 years old were evaluated. Foot deformity included 77 clubfeet of which 29 underwent Ponseti treatment and 48 underwent operative treatment. Seven feet were congenital vertical talus with 6 undergoing surgical intervention. At the time of gait analysis 11% of children wore KAFO's, 44% of children wore AFO's and 45% of children wore no orthotic or in shoe orthotic. Physical exam and kinematic data showed that children with arthrogryposis walked with a crouched gait, exhibited stiffness in the hips, knees, and ankles and showed limitations in their gross motor functioning ($p<0.006$). Power generation was low at the ankle and was high at the hip ($p<0.01$). Multi-segment foot kinematics revealed stiffness in hindfoot plantarflexion and residual forefoot adduction ($p<0.03$). Foot pressure showed reduced heel impulse, excessive midfoot contact, and overall varus foot position ($p<0.002$). Categorization by age revealed greater stiffness at the hips and knees in older children ($p<0.01$). Children with KAFO's showed the most stiffness ($p<0.05$). No significant differences were seen between the foot posture of children with clubfoot treated operatively or by Ponseti technique except for greater internal rotation at the ankle ($p=0.0158$) in Ponseti treated feet.

Table 1: Comparison of children with arthrogryposis as well as typically developing children for physical examination data, dynamic foot pressure data, kinetics data, and kinematics data. External rotation is negative. Forefoot adduction is negative. Varus is negative. SD: standard deviation. PROM: Passive Range of Motion.

Variable	Arthrogryposis		Normal value		*p*-value
	Mean	SD	Mean	SD	
Hip Abduction PROM (°)	34	13	45	5	<0.0001
Ankle Dorsiflexion PROM (°)	-4	15	9	7	<0.0001
Knee Extension PROM (°)	-10	16	4	4	<0.0001
Forefoot Ab/Adduction in physical exam (°)	-9	11	10	7	<0.0001
Varus-Valgus Position from foot pressure (°)	-34	52	3	30	<0.0001
Ankle Power Generation (W/kg)	0.99	0.6	2.7	0.4	<0.0001
Hip Extension in Stance (°)	-14	11	5	4	<0.0001
Knee Extension max in Stance (°)	-15	13	-5	2	<0.0001
Foot Rotation Angle Average in Stance (°)	13	12	-1	3	<0.0001
Gait Forward Velocity (cm/s)	84	23	117	12	<0.0001

DISCUSSION

Children with arthrogryposis walk with a crouched, stiff gait and have residual foot deformities even after treatment. Compensations for contractures and foot deformities are made to allow for a functional gait [5,6]. Due to stiffness in the ankles of children with arthrogryposis, push-off in walking is powered by the hips more than in typically developing children. Foot analysis revealed that clubfoot was the most common deformity in the population studied and recurrence of clubfoot deformity treated by either Ponseti or operatively was seen with varus foot posture and residual forefoot adduction. Quantification of contractures and deformities may help assess outcomes and guide treatment.

REFERENCES

1. Hall, J.G. (2014) Eur J Med Genet. Aug;57(8):464–72.
2. Carlson, W.O. et al. (2015 Foot Ankle Clin. Dec;20(4):619–44
3. Eriksson, M. et al. (2014) J Child Orthop. Aug;8(4):305–12.
4. Böhm, H. et al. (2013) Res Dev Disabil. Nov;34(11):4286–92.
5. Kroksmark, A-K. et al. (2006) Am J Med Genet A. Aug 15;140(16):1757–67.
6. Gutierrez, E.M. et al. (2005) Gait Posture. Jan;21(1):12–23.

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PARTITIONING GROUND REACTION FORCES FOR MULTI-SEGMENT FOOT KINETICS

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INTRODUCTION

Joint kinematics from multi-segment foot (MSF) models have been increasingly used in clinical gait analysis and human movement research. The addition of foot joint kinetics, however, has been limited, primarily due to technological limitations in measuring separate ground reaction forces (GRFs) under neighboring foot segments [1]. As the use of multiple force plates [1, 2] is impractical in most cases, a method that can accurately partition kinetics within the joints of the foot via a single force plate is attractive. Two such methods have been proposed. The first employs a pressure mat (secured to the top of the plate) to help partition the GRFs under each foot segment [3]. The second method quantifies joint kinetics from a single plate *only* when the location of the center-of-pressure (COP) passes anterior to a joint. This latter technique has been used to examine kinetics of the metatarsophalangeal (MTP) joint [4], but could theoretically also be applied to other joints in the foot. The purposes of the present study were to assess the accuracy of both methods on the calculation of midtarsal and MTP joint moments and powers. This was done by comparing their estimates to those obtained from a 'targeted foot placement approach' that isolated forces under two segments [5].

CLINICAL SIGNIFICANCE

A method that can accurately calculate foot joint kinetics from a single force plate will provide researchers and clinicians with a new tool with which to study intrinsic foot muscle function, better understand foot pathologies, and evaluate potential treatment interventions.

METHODS

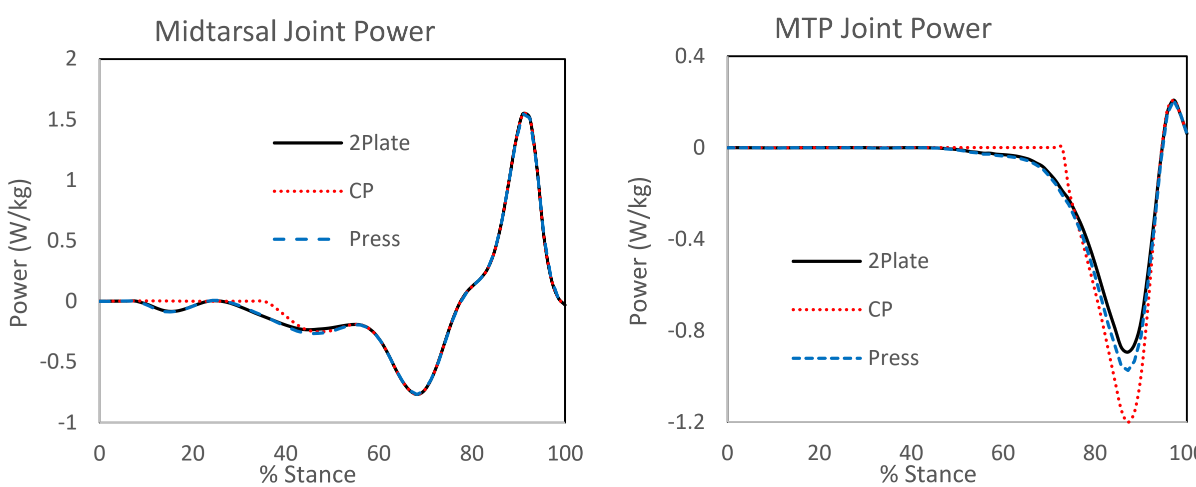
Nineteen reflective markers were placed on the right leg and foot of 13 healthy pediatric subjects (9M/4F; age 13.1 ± 3.1; height 156 ± 18 cm, weight 51 ± 18 kg) according to a previously described MSF model and collection protocol [5]. Briefly, subjects walked (using a three-step approach) across a floor containing 2 adjacent force plates so that the hindfoot contacted one plate while the forefoot contacted the adjacent plate. This was repeated with the hallux and toes on one plate and the rest of the foot on the adjacent plate.

Marker trajectories were collected at 120 Hz and filtered at 6 Hz, while GRF data was collected at 1560 Hz, filtered at 50 Hz, and threshold cutoff at 5N. Midtarsal (from the midfoot trials) and MTP (from the toe trials) joint moments and powers were calculated in Visual 3D software using three separate algorithms. In the first algorithm (Press), the normal forces and COPs for the two segments were taken from the individual force plates, mimicking the same partitioning when done using a pressure mat. The segment shear forces and free moments were then calculated from the combination of the two plates by assuming they were distributed in the same proportions as the normal forces [3]. In the second method (CP), the force plates were simply combined, and the entire GRF was applied to each segment sequentially based on the COP location (e.g. applied to the hindfoot (or foot) until the COP passed anterior to the midtarsal joint (or MTP joint), at which point it was applied entirely to the forefoot (or hallux)) [4]. The third algorithm (2Plate) used the separate GRFs from the adjacent force plates to calculate joint kinetics (e.g., moment, power and work) [5]. Joint kinetics for the Press and CP methods were calculated from the estimated, partitioned GRFs, and compared against 2Plate.

DEMONSTRATION AND SUMMARY

For brevity, only joint power profiles are presented (representative subject shown below). At the midtarsal joint, the Press method was nearly identical to 2Plate, while the CP method did not capture the first portion of power absorption. Negative joint work (integral of power absorption) was therefore underestimated by CP in 12 of the 13 subjects, by an average of 12%. Peak power absorption, as well as power generation, occurred entirely after the hindfoot left the ground and was therefore captured accurately by both methods. At the MTP joint, negative joint work was generally overestimated by both methods. Press averaged 5% overestimation, while CP errors ranged from slight underestimation (first portion not captured) to 115% overestimation (gross overestimation of the peak). Positive MTP joint work contained mostly minor errors, but considerable CP method overestimation occurred in two subjects (57% and 123%). Joint moments, while not presented, showed substantial errors for both methods in non-sagittal planes; however, their effect on joint powers was muted due to the dominance of the normal component of the GRF on joint power. Movements that exhibit, in particular, large M/L shear forces, may be more susceptible to errors from both methods.

Overall, midtarsal joint kinetics were captured well by both methods, primarily because the majority of midtarsal loading occurs when the COP is substantially anterior to the joint. MTP joint loading is distributed across the joint, creating greater error potential.



DISCLOSURE

The authors have no conflicts of interest to disclose.

REFERENCES

1. Bruening, D. et al. Journal of biomechanics, 2010. **43**(16): 3222-3226.
2. Yavuz, M. et al. Journal of biomechanics, 2007. **40**(13): 3045-3049.
3. Giacomozzi, C. et al. IEEE transactions on rehabilitation engineering, 1997. **5**(4): 322-330.
4. Stefanyshyn, D. et al. Journal of biomechanics, 1997. **30**(11): p. 1081-1085.
5. Bruening, D. et al. Gait & posture, 2012. **35**(4): 529-540.

A PEDOBAROGRAPH MEASURE TO ASSESS CENTER OF PRESSURE USING MOTION CAPTURE REGISTRATION

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INTRODUCTION

Plantar pressure analysis may be a useful tool for assisting with clinical decisions and outcomes; however, clinically meaningful and reliable metrics are lacking. In addition, in the absence of spatial registration of the foot to the plantar pressure, errors in interpretation of data may occur. To address these shortcomings a procedure utilizing motion capture coupled with the plantar pressure enables direct registration of the foot and pressure map. Using this registration method, a novel parameter to quantify the location and pattern of the center of pressure (COP) is introduced. This parameter is demonstrated to be both reliable and clinically meaningful over a spectrum of foot deformities.

CLINICAL SIGNIFICANCE

A plantar pressure based COP measure that accounts for the actual position of the foot using direct registration may allow analyses of pathologies to determine critical thresholds for intervention and inform clinical decisions.

METHODS

The method and measure were adopted for routine clinical analysis in August 2013. The collection method requires a set of retroreflective markers sufficient for establishing a single foot axis. Alternatively a validated foot model (mSHCG) may be used [1]. Motion data is captured simultaneous with plantar pressure. The plantar pressure mat is registered with respect to the lab coordinate system. The relationship between the subjects' COP and a typically developing (TD) cohort is assessed using the summation of the mediolateral (y) difference at each of 101 time normalized points over stance phase. Distance is normalized to marker based foot length and direction is established by a marker based foot axis. Both the sum, termed ML Index or MLI, and the absolute sum |MLI| are considered. The |MLI| is signed according to MLI to specify overall medial (positive) or lateral (negative) offsets.

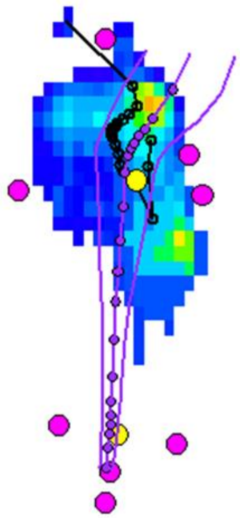
$$MLI = \sum_{n=1}^{101} (COPy_n - COP\_TDy_n)/101\,, |MLI| = sgn(MLI) \sum_{n=1}^{101} |COPy_n - COP\_TDy_n|/101$$

An IRB approved retrospective study was undertaken. The database was searched for studies using the mSHCG foot model back to August 2013. Both MLI and |MLI| were compared with a previously reported, unregistered COP measure, the center of pressure excursion index (COPEI) [2]. As the COPEI is purported to correlate with supination, this correlation was explored with all COP measures using the supination value reported from the mSHCG foot model during a single point (50%) of stance phase.

DEMONSTRATION

25 subjects (50 feet) were identified. Subjects represented a spectrum of deformities (data sample in Fig. 1). Mean (SD, Range) of MLI was -0.43 (±7.2, -19.5 – 16.4), |MLI| was -0.56

(±7.51, -19.5 – 16.4) and COPEI was 14.8 (±12.8, -15.9 – 45.2). Correlation between MLI and COPEI was $r^2 = 0.186$, |MLI| and COPEI was $r^2 = 0.176$. Bland-Altman analysis further indicated that these measures were not tightly coupled as two standard deviations of differences between measures ranged from -20 to +50, greater than the range of any of the individual variables. Correlation between MLI and supination was $r^2 = 0.710$, |MLI| and supination was $r^2 = 0.738$, and COPEI and supination was $r^2 = 0.134$.



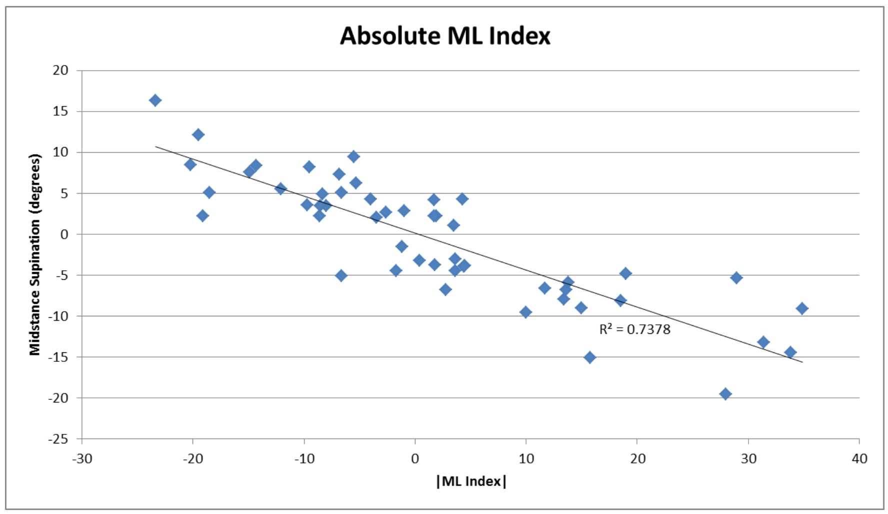


Figure 1. (Left) Example data illustrating markers (magenta circles), anterior and posterior foot axis references (yellow circles), subject COP (black) and control COP (purple). Figure 2. (Right) correlation of |MLI| with midstance supination from foot model.

SUMMARY

Registering a foot coordinate system with the plantar pressure map greatly increases both the accuracy and utility of COP measures[3,4]. Although a COPEI value could be obtained in all but one foot, it was clear based on the additional marker data that COPEI measures were inaccurate in about half of the feet used here. Using the registration method, all feet can be accurately assessed for MLI. COPEI relies on an axis established from a single posterior to an anterior COP point; in calcaneal or forefoot patterns this axis will not likely reflect the foot axis. Although both COPEI and MLI are dimensionless measures of the mediolateral COP offset, they were not well correlated. Additionally, only the MLI measures were strongly correlated with foot supination. The MLI considers the entire COP while the COPEI is determined by the two COP points used for the axis and one measured offset. The registration method requires little additional time and with further validation may provide useful data for clinical decision making.

REFERENCES: [1] Saraswat et al. Gait Posture 2012; 35:547–55. [2] Ledoux and Hillstrom. Gait Posture 2002; 15:1–9. [3] Giacomozzi et al. Med Biol Eng Comput 2000; 38:156–63. [4] Stebbins et al. Gait Posture 2005; 22:372–6.

DISCLOSURE STATEMENT: The author has no conflicts of interest to disclose.

Comparison of Three-Dimensional Multi-segment Foot Models

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INTRODUCTION

Many skin-mounted three-dimensional multi-segmented foot models are currently in use. Development and validation studies for these models have adequate procedures and sample selection, however, the methodical quality varies. Evidence regarding the repeatability, including between trial and between assessors, is mixed [1]. More importantly, there are no between model comparisons of kinematic results. Since each model utilizes different terminology and marker sets, and calculates local anatomical coordinate systems and intersegment rotations differently, there is no technical uniformity. This study examined five of these multi-segmented foot models: duPont [2], Heidelberg [3], Oxford Child [4], Leardini [5], and Utah [6]. While many share common marker placements, the types of angles that can be calculated as well as the difficulty in placing the markers consistently can vary. This study aimed to explore these differences in kinematics and repeatability between all five three-dimensional multi-segmented foot models in pediatric subjects.

CLINICAL SIGNIFICANCE

The results provide evidence based recommendations regarding the choice of foot models that can be used for clinical gait analysis and elucidate strengths and limitations of the choice.

METHODS

Individuals with no history of foot deformity or surgery were recruited for the study. To establish intra- and inter- physical therapist reliability, each subject underwent six marker placements; three placements by two PTs. An amalgamated set of 21 markers were placed on the feet and lower extremities. Marker positions were recorded by a twelve camera motion capture system (*Motion Analysis Co., Santa Rosa, CA).* A static calibration trial and five walking trials were collected for each subject.

Each model varies in the number of segments. All five models contain equivalent hindfoot, forefoot, and hallux segments. Dorsiflexion/Plantarflexion (DF/PF), varus/valgus (Var/Val), and internal/external rotation (IR/ER) angles were calculated for the hindfoot and forefoot segments, while DF/PF and IR/ER angles were calculated for the hallux segment (Figure 1). Average gait curves were calculated for each angle to evaluate overall patterns of motion. Root mean square (RMS) values were used to assess inter- and intra- rater reliability.

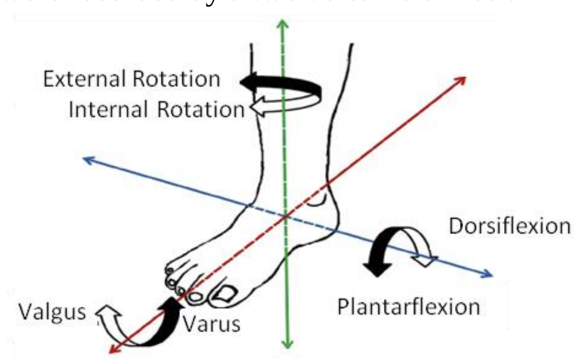

Figure 1: Foot Rotations

RESULTS

Ten individuals (Male = 6, Female = 4; Age = 19.8 ± 1.5) participated in the study. Offsets were present in the gait curves, yet all curves shared consistent patterns (Figure 2). The Heidelberg model produced significantly different kinematics than the other 4 models. The duPont and Oxford models were the most consistent with no significant differences seen between four of seven analyzed motion.

The duPont, Heidelberg, and Oxford models all had similar inter-rater RMS values (5.6, 5.9, and 6.0 respectively). The Utah model had an average RMS of 6.6. Leardini had the highest RMS values with an average of 11.2. The greatest variability in marker placement occurred in the rotational plane.

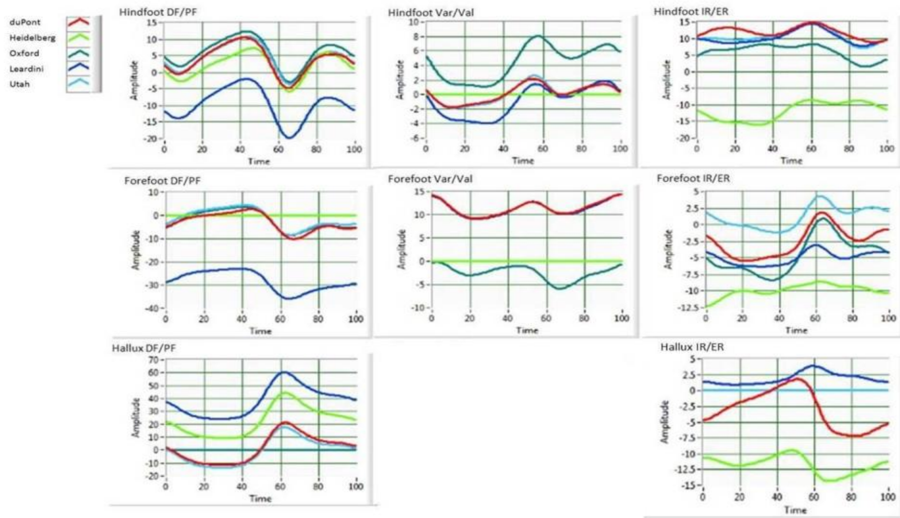

Figure 2: Gait Curves for all Measured Angles

DISCUSSION

The duPont and Oxford models shared the most kinematic similarities. Differences in angle calculation method made it difficult to directly compare the kinematics calculated with each model. While a significant offset was present, the overall kinematic pattern across a gait cycle was relatively consistent for most models. Four of the five models showed moderate repeatability with the exception being the Leardini model.

Given the similarities in kinematic data and relatively low variability, one model did not emerge as the preferred three-dimensional multi-segmented foot model. Rather, comparing findings between labs using the same model and using normative data for that specific model should allow for the appropriate clinical analysis to be made. One may want to consider the ease of marker placement and the investigator/physical therapist's familiarity with marker placement when choosing which model is appropriate for their lab or study.

REFERENCSE

1. Deschamps, K et al. (2011) Gait & Posture, 33(3): 338-349
2. Henley, J. et al (2008) Foot and Ankle Motion Analysis, pp. 445-463
3. Simon, J. et al. (2006) Gait & Posture, 23(4): 411-424
4. Stebbins, J. et al. (2006) Gait & Posture, 23(4): 401-410
5. Leardini, A. et al. (2007) Gait & Posture, 25(3): 453-462
6. MacWilliams, B.A. (2003) Gait & Posture, 17(3): 214-224

DISCLOSURE STATEMENT

The authors declare no conflicts of interest

Daily Schedule: Thursday 25 May 2017

GCMAS 2017 Annual Meeting

University of Utah Guest House and Officers Club

7:00A – 5:30P

TIME	TOPIC & ACTIVITY	ROOM
7:00A – 5:00P	Registration	Lobby
7:00A – 7:45A	Breakfast Session: Commission for Motion Analysis Accreditation (CMLA), Gordon Alderink, PT, PhD	Douglas Ball Room
8:00A – 5:00P	Exhibitors Open	Officers Club
8:00A – 9:30A	Podium Session 4 – Pathologic Gait/Prostheses I	Douglas Ball Room
9:30A – 11:00A	Exhibitors / Attended Posters - Even Numbered	Officers Club
11:00A – 12:00P	Keynote Speaker – Mike Aiona, MD	Douglas Ball Room
12:00P – 1:30P	GCMAS Business Luncheon OR Vendor Luncheon	Officers Club
1:30P – 3:00P	Podium Session 5 – Pediatric Gait / Cerebral Palsy II	Douglas Ball Room
3:00P – 4:00P	Exhibitor / Posters	Officers Club
4:00P – 5:30P	Podium Session 6 – Trunk and Upper Extremity	Douglas Ball Room
6:00P – 8:30P	GCMAS Banquet	Officers Circle Field

Breakfast Session

Commission for Motion Laboratory Accreditation (CMLA) Workshop

Wednesday, May 24th
7:00A - 7:45A

Presenter: Gordon Alderink, PT, PhD
President CMLA

Abstract: Each year, at the annual GCMAS annual, meeting the CMLA provides an update on the accreditation process and recognizes labs who have recently achieved full accreditation status. This year we want to have a “working” workshop.

During this early morning session CMLA Board members will provide an overview of accreditation, detailed explanation of the accreditation application, and use of the CMLA website. You are encouraged to bring working documents (including the applicant’s manual http://www.cmlainc.org/Portal.html) with you. The session will be informal, as you will get an opportunity to work one-on-one with Board members who have accreditation experience in their own labs and have served as application reviewers.

Anyone can attend this breakfast workshop, but those who are anticipating applying in the near future, currently working on re-submission, or beginning the application process will receive the most benefit. There is no fee, but registration is required.

For more information on the workshop contact Gordon Alderink at aldering@gvsu.edu OR about the CMLA and the accreditation process, please visit the website at www.CMLAinc.org.

Podium Session #4

PATHOLOGIC GAIT/PROSTHESES

MODERATED BY: **Marilynn Wyatt, PT:** Director, Gait Analysis Laboratory
The Geneva Foundation, Tacoma, WA

Audrey Zucker-Levine, PhD: School of Physical Therapy
University of Saskatchewan, Saskatchewan, Canada

1. **Bilateral Gait Mechanics in Unilateral End-stage Ankle Osteoarthrosis**
   Swati Chopra, Xavier Crevoisier

2. **Muscle Activation Patterns During Walking Are Correlated to Clinical Gait Assessments After Traumatic Brain Injury**
   Samuel Acuna, Mitchell Tyler, Yuri Danilov, Darryl Thelen

3. **Cognitive Control of Tandem Walking in Middle Aged to Older Adults with Multiple Sclerosis**
   Manuel Hernandez, Gioella Chaparro, Roee Holtzer, Meltem Izzetoglu, Robert Motl

4. **Gait Parameters of Patients with Parkinson Disease to Develop Innovative Virtual Reality Training Environments**
   Margaret Patterson, K. Bo Foreman, Andrew Merryweather

5. **Gait, Balance, And Coordination Impairments in Niemann-Pick Disease, Type C1**
   Ashwini Sansare, Cris Zampieri, Katherine Alter, Christopher Stanley, Nicole Farhart, Lee Ann Keener, Simona Bianconi, Forbes Porter

6. **Gait Outcomes Post Proximal Tibial Tumor Resection and Endoprosthetic Reconstruction**
   Andy Vuong, Kent Yamaguchi, Kent Heberer, Marcia Greenberg, Nicholas Bernthal, Jeffrey Eckardt, Eileen Fowler

7. **Vacuum Level Effects On Knee Contact Force for Unilateral Transtibial Amputees with Elevated Vacuum Suspension**
   Hang Xu, Kasey Greenland, Donald Bloswick, Andrew Merryweather

8. **Kinematics Comparison of Early Knee Versus Traditional Knee Prosthetic Prescription in Young Children with Amputation**
   Mark Geil, Zahra Safaeepour

Bilateral Gait Mechanics in Unilateral End-stage Ankle Osteoarthrosis

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INTRODUCTION

Gait asymmetry following ankle surgeries has been regularly reported [1], however, the relations between pre and postoperative compensatory gait patterns are not clear. This study assesses the bilateral gait mechanics of ankle osteoarthrosis (AOA) patients in order to further understand compensatory gait alterations.

CLINICAL SIGNIFICANCE

- Intersegment coordination patterns could add more qualitative information to kinematic assessment in patients with AOA and could prove helpful in rehabilitation planning.
- A feasible and less expensive ambulatory gait analysis system makes it possible to test patients in an open environment, capturing data from several consecutive gait cycles.
- A more in-depth understanding of the compensatory gait patterns of patients with AOA may help predetermine the extent of prognosis.

METHODS

20 participants, including 10 controls (age 64.9±9.1, BMI 25.7±5.5) and 10 patients with end-stage AOA (age 65.8±8.9, BMI 27.6±3), were assessed using 3-D inertial sensors Physilog® and pressure insoles PEDAR®. 46 gait parameters were studied including 10 spatiotemporal, 6 kinematic, including range of motion (RoM) and intersegment coordination (ISC) using continuous relative phase (CRP) in the sagittal plane [2], and 30 plantar pressure parameters (PPP), including total contact duration, maximum force (%BW) and maximum pressure (kPa) in 10 foot subregions (hindfoot and midfoot (lateral and medial), forefoot (lateral, central and medial) and toes (first, second and lateral)). The full protocol can be found in a previous publication [1]. Assessed parameters in the patient group were compared both bilaterally and with controls.

RESULTS

Spatiotemporal outcome suggests an alteration compared to the controls, but a generally good bilateral symmetry (Table 1). Kinematic results show a restricted RoM of the affected side, with ISC reporting an increased compensatory forefoot rotation (Fig 1). PPP report bilateral asymmetry across all three toe subregions. However, in comparison to the controls, both sides of the AOA group report a reduced loading ($p<0.05$) in the hindfoot medial subregion at the point of initial contact. Furthermore, the unaffected side of AOA patients also reported reduced contact forces at the forefoot medial and central subregions.

DISCUSSION

The combined outcome of 46 parameters suggests that AOA patients walk with an altered gait pattern. However, is it of note that the previous assumption that the unaffected side duly compensates by an increased loading, as is the case with other weight bearing joints, is not completely true. This study shows that the affected side compensates for the insufficiencies at the ankle joint, predominately due to pain and restricted motion. The compensation strategy also includes a reduced toe loading, further reducing push-off and peak swing speed to optimize walking with the reduced ankle RoM. Postoperatively, this strategy is not reported after total ankle replacement but is seen to worsen following ankle arthrodesis [1].

In conclusion, this study suggests that the post-operative gait alteration and asymmetry is not entirely due to the preoperative gait pattern, but rather it is due to the type of surgery and subsequent rehabilitation programs which are the major contributing factors.

Table 1: Spatiotemporal parameter of gait, result in mean (SD)

Spatiotemporal parameters	AOA affected side	AOA unaffected side	Control	ANOVA p value
Stance (GCT%)	58.67 (2.2)	59.79 (3.4)	59.3 (2.4)	0.63
Cadence (steps/min)	99.3 (13.6) †	98.7 (14.8) †	111.3 (8.9)	0.03
Load (St%)	9.7 (2.7)	8.9 (2.8)	10.7 (1.9)	3.32
Foot-flat (St%)	60.2 (10.2)	61.7 (10.1) †	55.9 (4.6)	0.03
Push-off (St%)	30.0 (9.7)	29.4 (9.3) †	33.5 (4.7)	0.04
Stride (m)	0.99 (0.26) †	1.0 (0.25) †	1.26 (0.1)	0.014
Speed (m/s)	0.84 (0.3) †	0.85 (0.3) †	1.18 (0.13)	0.006
Peak swing speed (°/s)	290.5 (77.6) †	295 (69.4) †	385.7 (49.3)	0.004
Toe-off pitch angle (°)	-55 (10.8) †*	-59.2 (10.7) †	-69.35 (4.86)	0.003
Heel strike pitch angle (°)	16.4 (5.98)	15.59 (5.7)	19.5 (3.59)	0.22

† Significantly different to the controls, * significant difference between the two sides ($p<0.05$)

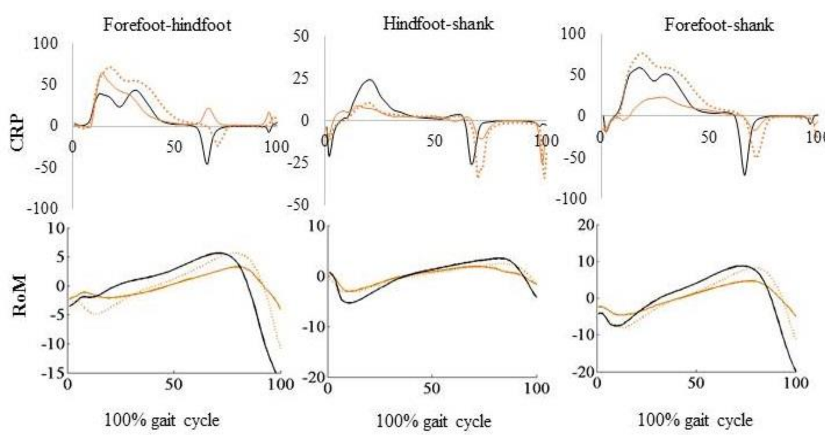

Figure 1: CRP and RoM in three intersegments for controls (black), AOA affected (orange solid) and AOA unaffected (dotted). All three intersegments showed significant difference on the affected side of AOA compare to their unaffected sides or to the controls ($p<0.05$).

REFERENCES

1. Chopra, S. et al. (2014) Journal of Orthopedic Research, 32(3):377-84.
2. Chopra, S. et al. (2016) Journal of Orthopedic Research, doi: 10.1002/jor.23379.

DISCLOSURE STATEMENT

No conflicts of interest to disclose.

MUSCLE ACTIVATION PATTERNS DURING WALKING ARE CORRELATED TO CLINICAL GAIT ASSESSMENTS AFTER TRAUMATIC BRAIN INJURY

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INTRODUCTION

Chronic gait deficits are often present in persons who have sustained a mild to moderate traumatic brain injury (mTBI)—they tend to walk more slowly, take shorter steps and exhibit greater mediolateral sway [1]. Gait deficits in mTBI have been assessed clinically using the Dynamic Gait Index (DGI) [1–4], an eight task battery (score: 0-24) that scores an individual's ability to walk normally, navigate obstacles, and turn. DGI has been established as a valid predictor of falls in mTBI [3]. While clinically useful, DGI cannot provide insights into the underlying disruption in neuromuscular control that may give rise to abnormal gait. Quantitative electromyographic (EMG) analysis could potentially be used to identify disruptions in normal phasing of muscle activation patterns. The primary objective of this study was to investigate the relationship between muscle activation during walking and DGI assessments in a cross-section of individuals with chronic gait abnormalities due to mTBI. A secondary objective was to assess whether gait EMG patterns are related to dynamic posturography metrics in mTBI.

CLINICAL SIGNIFICANCE

Traumatic brain injury can permanently disrupt normal brain function and result in long term deficits in physical function. In this study, we show a moderate relationship between clinical gait assessments and an objective measure of the disruption of normal EMG patterns during walking. These results suggest that quantitative EMG analysis could potentially be useful to track subtle changes in the coordination of gait that may arise with mTBI and neurorehabilitation.

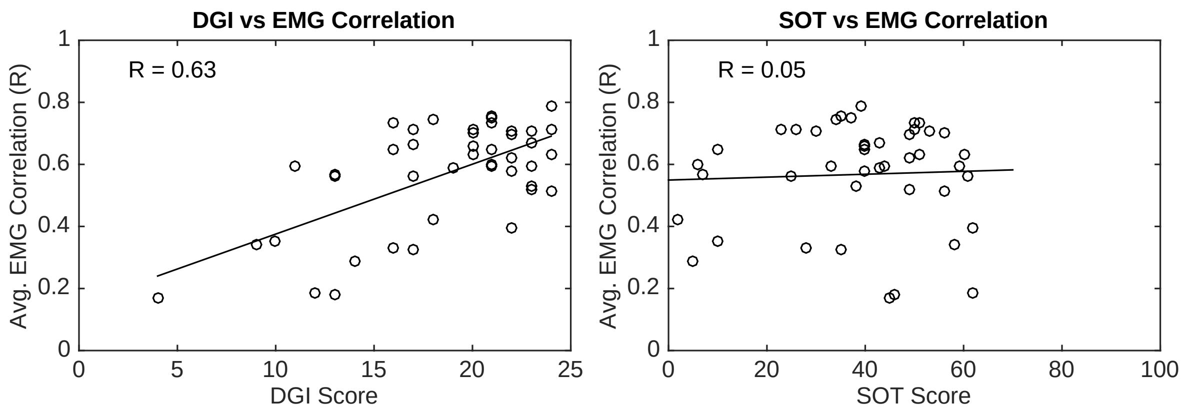
METHODS

We enrolled 44 ambulatory adults (age 18-65) presenting a chronic balance deficit due to mTBI. All subjects were at least one year post-injury, and scored at least 8 points below normal on the sensory organization test (SOT, NeuroCom®), an objective measure of standing balance under six sensory conditions that challenge visual, vestibular and proprioceptive systems. The DGI was administered by trained physical therapists.

Lower extremity electromyographic (EMG) signals were collected during treadmill walking (60 seconds) at each subject's baseline preferred speed (2.2±0.5 mph). Surface electrodes were placed over six muscles on each leg (Tibialis Anterior, Medial Gastrocnemius, Soleus, Vastus Lateralis, Rectus Femoris, Semitendinosus). EMG signals were rectified and then low-pass filtered at 10 Hz. Ensemble EMG patterns were created by averaging signals over all gait cycles and normalizing the data to the muscle's root-mean-squared EMG. The protocol was repeated on 20 healthy young adults (10 F, age 25±3, 2.2 mph). We cross-correlated individual mTBI muscle activation patterns with the healthy control EMG data to assess how mTBI muscle activations differed from normal [5]. The average correlation coefficient for each subject was then compared to their DGI and SOT scores using a Pearson's correlation.

RESULTS

DGI scores for the mTBI subjects averaged 18.5±4.6, putting most subjects at an increased risk for falls [3]. SOT scores averaged 39±17. The average cross-correlation of mTBI and normative EMG data was 0.57±0.16. There was a strong linear relationship between the DGI and the EMG ($R=0.63$, $p<0.01$). However, we found no relationship between the SOT and EMG correlation ($R=0.05$) over the mTBI subjects.



Figures 1, 2: For 12 leg muscles, EMG activation patterns over the gait cycle were generated for mTBI subjects and healthy controls. The mTBI patterns for each muscle were cross-correlated with the healthy patterns. The vertical axes show the average resulting correlation coefficient. The Dynamic Gait Index (DGI) is correlated with this coefficient ($R = 0.63$), but the Sensory Organization Test (SOT) is not correlated ($R = 0.05$).

DISCUSSION

It has previously been shown that average cross-correlations of individual EMG patterns between different healthy subjects range from 0.6 to 0.9 [5]. The mTBI subjects in this study had average cross-correlations across all muscles ranging from 0.2 to 0.8, suggesting a broad range of disruption in normative muscle activation patterns present in the subjects tested. The moderate relationship between EMG and DGI is promising, and motivates future work to determine if quantitative EMG analysis can identify specific disruptions in coordination patterns and track subtle, but potentially important, improvements in neuromuscular control that can arise with neurorehabilitation. Synergy analysis could provide a promising next step to characterize the inter-relationship among activation patterns that may exist in cyclic motor tasks such as gait [6].

REFERENCES

1. Basford, J. et al. (2003) Arch. Phys. Med. Rehabil., 84(3):343-349.
2. Kaufman, K. et al. (2006) Med. Eng. Phys., 28(3):234-239.
3. Medley, A. et al. (2006) Brain Inj., 20(13-14):1403-1408.
4. Williams, G. et al. (2009) Arch. Phys. Med. Rehabil., 90(4):587-593.
5. Wren, T. et al. (2006) J. Biomech., 39(14):2714-2718.
6. Steele, K. et al. (2015) Dev. Med. Child Neurol., 57(12):1176-1182.

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DISCLOSURE STATEMENT

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COGNITIVE CONTROL OF TANDEM WALKING IN MIDDLE AGED TO OLDER ADULTS WITH MULTIPLE SCLEROSIS

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INTRODUCTION

Gait impairments are common in persons with multiple sclerosis (MS), and are expected to worsen with increasing age. MS is typically diagnosed between 20 and 50 years of age, but as the peak prevalence of MS shifts into older age groups [1], these individuals are likely to experience a combination of both age-related and MS-related declines in gait function. Clinical walking tests such as the 3-meter Timed Tandem Walking Test have been found to detect impairment of the motor or cerebellar system in persons with MS that complements the well-established Timed 25-Foot Walk [2]. Recent findings suggest that during simple and complex walking tasks, increased attention might be required by persons with MS relative to healthy controls [3]. However, no studies to date, in part due to limitations of conventional neuroimaging methods, have examined the cognitive control involved in coordinating precise foot placements while walking. Furthermore, the study of gait impairment in persons with MS is clinically relevant, particularly in complex tasks that might better represent the demands of ambulation in the community.

This study examined the cognitive control of tandem walking in MS using functional near-infrared spectroscopy (fNIRS), an emerging neuroimaging technique for evaluating cortical activation during walking and cognitive tasks. Based on prior findings of increased activation levels of the pre-frontal cortex (PFC) during walking and talking compared to walking alone in older adults and in increases of activation levels in persons with MS relative to controls [3,4], we hypothesized that in comparison to healthy controls (HC group), middle-aged to older adults with MS (MS group) would exhibit increased PFC activation levels while walking at a similar gait speed. Secondly, we hypothesized that as the complexity of the walking task increased, PFC activation levels would increase across all groups.

CLINICAL SIGNIFICANCE

The use of functional near infrared spectroscopy while walking may provide an effective tool in quantifying changes in the cognitive control of gait in older adults with MS.

METHODS

Participants. Ten middle-aged to older adults with MS (8 Female; Mean [SD] Age = 53±5 years; Body Mass Index [BMI] = 25±4) with an expanded disability status scale (EDSS) score of 4±2, and 12 healthy controls (HC, 9 F, 63±4 years old, BMI = 27±6) were recruited from the local community.

Procedure. A 3-m instrumented treadmill was utilized for all walking conditions. Participants performed three self-paced treadmill walking tasks while wearing a ceiling-mounted harness for safety and an fNIRS sensor attached to their forehead: normal walking (NW), straight-line walking (SLW) and tandem walking (TW). For SLW, participants were further instructed to walk on a straight line without touching their heel to toe. For TW, participants were further instructed to walk by touching heel to toe. Measures collected included spatiotemporal gait parameters (i.e., stride length, step width, and gait speed) and PFC activation levels, evaluated by oxygenated hemoglobin (HbO_2) levels. The effect of group (MS vs. HC) and task (NW, SLW vs. TW) on outcome measures was evaluated using linear mixed models, with intercepts for individuals modeled as a random effect. Significance was set at $p = .05$, using R 3.1.1.

RESULTS

Overall, MS demonstrated a 23% increase in PFC oxygenation levels, when compared to HC (p=.02), while achieving similar gait speed, step widths, and stride lengths (p>.05). As hypothesized, a significant 5-fold and 9-fold increase in PFC oxygenation levels was observed in SLW and TW tasks relative to NW (p <.001). Furthermore, a significant interaction effect between group and task was observed (Fig. 1, p=.02), such that HC, with an increased availability of cognitive resources had higher relative changes observed in HbO_2 levels when going from NW to SLW and TW.

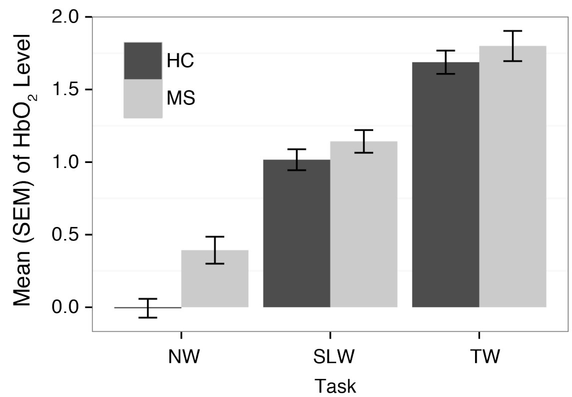

Figure 1. Mean PFC oxygenation level changes during normal walking (NW), straight-line walking (SLW) and tandem walking (TW) on a self-paced treadmill in persons with multiple sclerosis (MS) versus healthy controls (HC).

DISCUSSION

The increases in PFC activation levels observed in persons with MS relative to healthy controls during self-paced treadmill walking is consistent with prior observations in over ground walking [3]. Furthermore, the increases in HbO_2 levels observed from NW to SLW, and SLW to TW suggest increased attention is necessary to control precise foot placement in TW, above the attentional demands required for the maintenance of balance due to mediolateral step restrictions. Future studies should examine a larger sample of age-matched individuals and explore the relationship between other aspects of gait performance, such as temporal and spatial variability and changes in attentional demands.

REFERENCES

1. Marrie, RA. et al. (2010) Neurology, 74: pp. 465-471.
2. Stellmann, J.P. et al. (2014) Journal of the Neurological Sciences, 346: pp. 99-106.
3. Hernandez, M.E. et al. (2016) Journal of the Neurological Sciences, 370: pp. 277-283.
4. Holtzer, R. et al. (2015) Neuroimage, 112: pp. 152-159.

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DISCLOSURE STATEMENT

All authors have no conflicts of interest to disclose.

GAIT PARAMETERS OF PATIENTS WITH PARKINSON DISEASE TO DEVELOP INNOVATIVE VIRTUAL REALITY TRAINING ENVIRONMENTS

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INTRODUCTION

Parkinson disease (PD) is a decline of the central nervous system that mainly affects the motor system. Decline of the motor system often causes a stooped posture, tremors, and unbalanced gait [1]. The unbalanced gait associated with PD coupled with the general consequences of aging, increase falls while walking by 50% [2] and complicate day to day dual task (DT) activities such as walking while talking and stepping up onto a curb. To reduce falls and make DT activities easier for individuals with PD, innovative therapy interventions such as virtual reality (VR) training environments show promise. These environments allow patients to walk in the presence of multiple stimuli in a safe and controlled environment [3]. The relationship between a DT and gait in PD needs to be better characterized so comprehensive and effective VR training environment can be developed.

CLINICAL SIGNIFICANCE

Understanding changes in gait that a patient with PD makes when a cognitive task and physical task are introduced we can make the case for a more comprehensive virtual reality training system that will facilitate improved training in a safe and controlled environment.

METHODS

The goal of this study was to determine how a cognitive dual task (serial seven subtractions) affects the gait of patients with PD while performing an obstacle crossing task. The influence of a cognitive dual task activity on gait in patients with PD has been studied by others [2], yet the effects of VR training environments on gait parameters in PD are not well defined. In this study, data from 10 patients with mild to moderate PD and 10 age matched controls were recorded as they walked on an instrumented treadmill and performed a combination of tasks; walking only and walking while crossing over an obstacle, both with and without the DT. Spatiotemporal parameters of double limb and single limb support time, stride length, and steps per minute were quantified with Vicon Nexus (Vicon, Lake Forest, CA) and Visual 3D (C-Motion Inc., Germantown, MD). Single paired t-tests (α=0.05) were performed on the means of the gait parameters. Walking trials were compared to the matched DT trials to determine variability in spatiotemporal parameters when a cognitive dual task is introduced.

RESULTS

Overall patients with PD showed a significantly shorter, faster gait than their control counterparts. Within PD there was no significant difference between parameters when the secondary cognitive task of a serial seven subtraction was added to the walking and obstacle trials ($p>0.05$). While the PD group had significantly smaller single limb support ($p=.0039$) the double limb support between the two groups was not significant. The average steps per minute of the PD group is significantly greater than the controls ($p=0.02$). The average stride length of PD is significantly shorter than the controls ($p=.0008$). The ratio of single and double limb support was computed (Figure 1). The change in double limb support time for PD was unexpected.

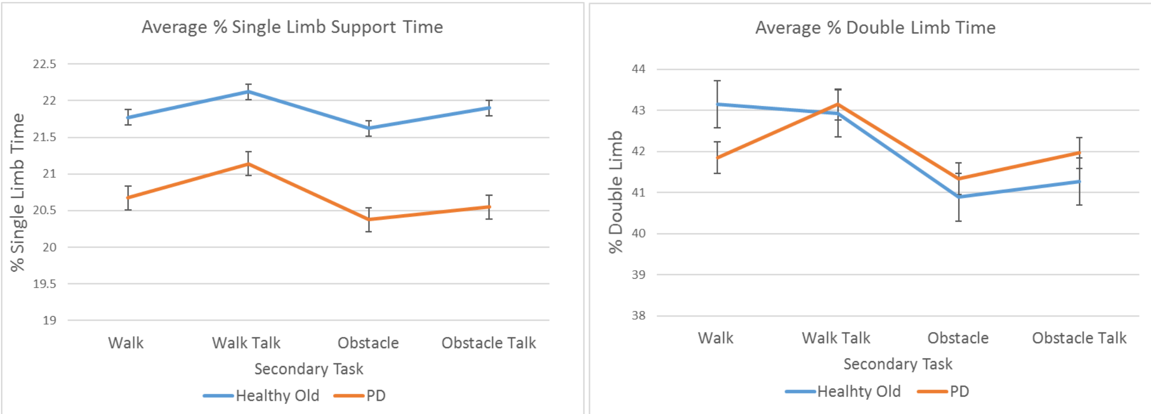

Figure 1- Gait Parameters. (Left) The average single limb of PD is significantly shorter than the controls ($p=0.039$). (Right) The average double limb of PD is not significantly different ($p>0.05$).

DISCUSSION

The shorter, faster stride length and decreased variability within PD suggest that gait adaptations and compensatory strategies differ between groups. Smaller single limb support and stride length, coupled with greater steps per minute are indicative of a shuffling gait associated with PD. All gait parameters in PD varied, but not significantly between conditions while the controls had greater variability between conditions. This may indicate that PD prioritized walking due to a fear of falling. These results differ slightly from other published values [2], most likely attributed to the treadmill testing environment which forced a set speed for all participants. This suggests that enhanced VR training environments could take advantage of the large adaptation between walk and obstacle conditions to provide terrains and obstacles that are difficult or impossible to create in other non-VR environments. VR gait training also offers greater control of the environment while maintaining a sense of safety for the patients, creating a positive training experience. In summary, the adaptations observed in this study suggests that treadmill training alone may not be sufficient to train adaptability during gait. Enhanced VR training environments to influence gait adaptations while training should be explored.

REFERENCES

1. O'Shea, M. et al. (2002) Physical Therapy, 82, pp. 888-896.
2. Eusterbrick, A.J. et al. (2011) Parkinson's disease, 2012, pp. 1-14.
3. Mirelman, A. et al. (2010) J Gerintil A Biol Sci Med Sci, pp. 1-7.

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DISCLOSURE STATEMENT

The authors declare no conflicts of interests to disclose.

GAIT, BALANCE, AND COORDINATION IMPAIRMENTS IN NIEMANN-PICK DISEASE, TYPE C1

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INTRODUCTION

Niemann-Pick disease type C1 (NPC1) is a lysosomal storage autosomal recessive disorder with an estimated incidence of approximately 1/100,000-120,000 live births [1]. The pathophysiology involves lipid accumulation in various tissues due to impaired lipid transport leading to a broad spectrum of neurovisceral manifestations. The typical neurological manifestations include progressive cerebellar ataxia, dystonia, dysarthria, dysphagia, and dementia. The resulting gait, balance, and coordination abnormalities may present clinically as slow and guarded movements, frequent falls, unsteadiness and support seeking tendency during ambulation, clumsiness, and difficulty with fine motor skills.

CLINICAL SIGNIFICANCE

Although objective gait, balance, and coordination measures have been extensively reported in other neurological pediatric populations, there is a dearth of literature that quantifies and objectively assesses these deficits in NPC1, using valid and reliable methods. The purpose of this study is to objectively assess gait, balance, and upper limb coordination in the NPC1 patients in comparison with healthy age and gender-matched controls. This study is a major step towards better understanding the functional limitations in the NPC1 population and finding potentially sensitive measures to track disease progression and treatment effectiveness.

METHODS

A cross-sectional analysis was performed between 10 patients with NPC1 (4-23 years old) and 10 healthy age and gender-matched controls. Spatiotemporal gait parameters were collected with either the GaitRite® Walkway System or Vicon® Motion Analysis System [2]. The balance measure of choice was sway velocity during quiet stance, in all four conditions (eyes open-EO and closed-EC, firm, and foam surfaces) of the modified Clinical Test of Sensory Interaction on Balance (mCTSIB) on the NeuroCom SMART EquiTest® System [3]. The classic clinical finger-to-nose test was quantified using the Vicon® Motion Analysis System [4] to assess upper limb (UL) coordination. Path length and path time of the index finger were calculated over three repetitions of the finger moving between the patient's nose and a target placed on a table in front of them. Between-group comparisons were conducted using a t-test for normally distributed variables, and Man-Whitney U tests for non-parametric variables.

RESULTS

Table 1 shows demographic characteristics of the NPC1 group, along with a checklist identifying the tests of balance, coordination and gait that the patients were able to complete. Due to varying levels of impairment, not all patients were able to perform all tests. The most common limitations during testing among the patients were: 1) non-ambulatory, 2) too young or cognitively challenged, leading to difficulty following directions. There were statistically significant differences between the two groups on the first two conditions of mCTSIB and all the UL motion analysis parameters (Table 2). On the mCTSIB, patients swayed faster than controls during quiet stance with EO and EC on firm surface. On the finger-to-nose task, patients took longer to complete the task, displaying a longer path compared to controls. During gait analysis, all the parameters except cadence, step length, and step width were significantly different between groups; patients walked slower while taking longer to cover the same distance, and spent more time on the double support phase of walking. Furthermore, patients showed more variability in step time and step length compared to controls.

Table 1: Patient characteristics and test adherence. Checkmarks indicate the patient was able to complete the tests.

ID	Age	Sex	mCTSIB	UL Reaching	Gait
1	4.7	F		✓	
2	8.0	M	✓	✓	✓
3	10.7	F	✓		✓
4	13.9	F	✓	✓	✓
5	16.5	F	✓	✓	✓
6	18.0	M	✓	✓	✓
7	18.4	F			✓
8	20.1	F	✓	✓	✓
9	22.6	F		✓	✓
10	23.4	M	✓	✓	

Table 2: Statistically significant balance, UL reaching and gait parameters on inter-group comparison. mCTSIB conditions on Foam EO and EC were not reported because most patients were unable to perform these tests, yielding n<5, therefore not included in statistical tests.

	mCTSIB		UL Reaching				Gait				
	FirmEO	FirmEC	R PLn	LPLn	R TT	L TT	Vel	ST	CoVrSLn	CoVrST	DSupp %
p value	0.038	0.035	<0.001	<0.001	0.034	0.012	0.011	0.015	0.003	<0.001	0.003

Vel = Velocity, ST = Step Time, CoVrSLn = Covariability Step Length, CoVrST = Covariability Step Time, DSupp% = Double Support %, RPLn = Right Path Length, LPln = Left Path Length, RTT = Right Task Time, LTT = Left Task Time.

DISCUSSION

This is the first reported study to objectively measure motor function in children diagnosed with NPC1. Our findings show that several of the gait parameters, and all of the balance and UL parameters statistically analyzed in this investigation were sensitive to differentiation between children with NPC1 and healthy controls. Obtaining statistically significant results with a small and variable sample of patients is encouraging for the research and clinical fields. Our findings support further investigation of these variables as potential markers of disease progression on motor function, as well as potential parameters for testing the effectiveness of interventions in this population.

REFERENCES

1. Patterson, M. et al (2013). Orphanet Journal of Rare Diseases, 8(12) 1-10.
2. Thorpe, D. et al (2005) Archives of Physical Medicine and Rehabilitation, 86(12) 2342-46.
3. Geldhof, E. et al (2006) European Journal of Pediatrics, 165, 779-786.
4. Jaspers, E. et al (2011) Research in Developmental Disabilities, 32(6), 2283-2294.

DISCLOSURE STATEMENT

The authors have no conflicts of interest to disclose.

Gait outcomes post proximal tibial tumor resection and endoprosthetic reconstruction

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INTRODUCTION

Endoprosthetic reconstructions after orthopedic tumor resections are the current standard of care [1]. The majority of outcome studies have focused on subjective outcomes rather than objective functional data [1]. The few studies that report gait kinematics and kinetics have focused on proximal and distal femur tumor resections and there is very little information about proximal tibial tumor resection (PTR) and endoprosthetic reconstruction [2][3]. Post PTR, there is potential for functional loss due to extensive excision of the proximal tibia including the patella tendon and ligaments as well as a medial gastrocnemius transfer to reconstruct the extensor mechanism. In this study, hip, knee and ankle kinematics and kinetics at self-selected and fast walking speeds were assessed for individuals post PTR. We aimed to evaluate functional loss of the surgical limb and compensatory mechanics of the non-surgical limb compared to a control group of similar age and gender. The inclusion of fast walking speeds may reveal functional loss not apparent at self-selected speeds.

CLINICAL SIGNIFICANCE

The outcomes of this study may provide patients and surgeons with reasonable expectations for walking ability post endoprosthetic reconstruction.

METHODS

Nine participants who were at least 2 years post PTR and over the age of 18 years as well as nine control subjects of similar age and gender were recruited. Three-dimensional gait analyses were performed during barefoot independent walking using a modified Helen Hayes marker set, an Eagle eight-camera system (Motion Analysis Corporation, Santa Rose, CA) and Cortex 1.1 (Motion Analysis Corporation). For fast speeds, participants were asked to walk as quickly and safely as possible. Sagittal plane kinematics and kinetics of the hip, knee and ankle joints were calculated in Orthotrak 6.6.1 (Motion Analysis Corporation), with the kinetics data normalized to body weight. Critical points for the angular position, moment and power throughout a gait cycle were identified for these joints [4]. Averaged gait data for the surgical and non-surgical limbs were compared to that of the control group for both self-selected and fast walking speeds. Independent t-tests were performed to determine statistically significant differences with significance set at $p<0.05$.

RESULTS

The mean age was 31 years for both the PTR and control groups with similar age ranges of 18-42 years and 18-44 years, respectively. Each group also had four males and five females.

Kinematics and kinetics of the PTR group's lower limbs at self-selected speeds can be seen in Figure 1. Compared to the control group, the surgical limbs showed an absence of knee flexion during early stance (K2, $P<0.001$), excessive peak plantarflexion during early stance (A2, $P<0.001$) and excessive peak dorsiflexion during late stance (A3, $P<0.001$). There was inappropriate knee flexor moment during early stance (KM1, P<0.001) and reduced peak plantarflexion moment (AM2, P=0.004). The PTR group's non-surgical limbs exhibited excessive peak knee flexion during early stance (K2, P=0.02) as compared to the control group.

Fast walking speed comparisons are shown in Figure 2. At faster speeds, the control subjects exhibited higher values for all critical points except A2, which was lower at preferred walking speeds. This change resulted in more pronounced kinematics and kinetics gait deviations for the surgical limb compared to the controls for K2, A2, KM1, KP1 (peak knee power absorption during early stance) and KP2 (peak knee power generation during early stance). Reduced peak hip extension (H3, P=0.04), reduced knee flexion at initial contact (K1, P=0.03) and excessive peak hip joint power during early stance (HP1, P<0.001) were also observed. The non-surgical limbs at fast speeds showed similar compensatory mechanisms at self-selected and fast speeds with greater deviations for KP2 only.

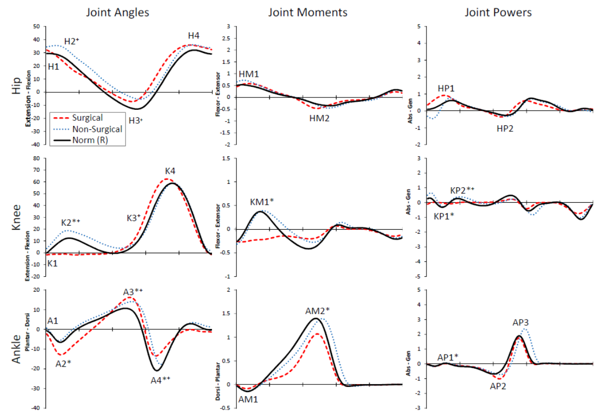

Figure 1: Kinematics and kinetics at self-selected speeds. *P<0.05 for surgical limb. $^{+}$P<0.05 for non-surgical limb compared to controls. Data are plotted as 100% of the gait cycle from initial contact of the foot with the floor to the next initial contact.

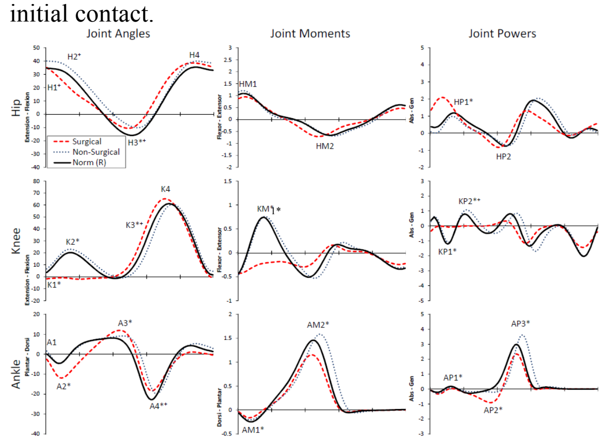

Figure 2: Kinematics and kinetics at fast speeds. *P<0.05 for surgical limb. $^{+}$P<0.05 for non-surgical limb compared to controls. Data are plotted as 100% of the gait cycle from initial contact of the foot with the floor to the next initial contact.

DISCUSSION

To our knowledge, this is the first study to focus on gait outcomes post PTR. For the surgical limbs at self-selected speeds, significant gait deviations were found for both the surgical and nonsurgical limbs, with additional and more pronounced deviations at fast speeds. These deviations are likely due to significant knee extensor weakness previously reported for patients post PTR [1].

REFERENCES

1. Bernthal, N. M. et al. (2014) Clinical Orthopedics and Related Research, 473(3): 812–819.
2. Kawai, A. et al. (2000) The Journal of Bone and Joint Surgery, 82(5): 666–669.
3. Kawai, A. et al. (1999) Journal of Surgical Oncology, 70(2): 109–115.
4. Okita, Y. et al. (2013) Clinical Biomechanics, 28(8): 898–903.

DISCLOSURE STATEMENT

The authors have no conflicts of interest to disclose.

VACUUM LEVEL EFFECTS ON KNEE CONTACT FORCE FOR UNILATERAL TRANSTIBIAL AMPUTEES WITH ELEVATED VACUUM SUSPENSION

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INTRODUCTION

The elevated vacuum suspension system (EVSS) can provide a better fitting socket and a superior linkage between the residual limb and prosthesis. Previous studies report a reduction of vertical pistoning in the EVSS socket, an increase in rotational stability and reduction in volume loss of the residual limb [1-2]. Even with all these known benefits, little attention has been paid to the effect of different vacuum pressure settings including what should be considered a sufficient or preferred vacuum level. Therefore, the objective of this study was to use 3D muscle-driven gait simulation to investigate the effect of vacuum levels on knee contact force (KCF) in both the residual and intact limbs for unilateral transtibial amputees (TTA).

CLINICAL SIGNIFICANCE

The knowledge gained could benefit amputees, clinicians and prosthetic designers by providing a better understanding of the effect of vacuum level on gait, and possibly help prevent early onset of knee osteoarthritis and influence EVSS design.

METHODS

Kinematic and force plate data of nine male unilateral TTA using the EVSS and nine non-amputees were analyzed. Institutional review board approval was obtained. The participants walked over ground at a controlled speed (1.20 ~ 1.40 m/s). Five trials were collected for unilateral TTA at each vacuum level (0 to 20 inHg, 5 inHg increments) and the non-amputees.

An OpenSim model which included 23 degrees of freedom (DoF) and 54 muscles was used for analysis [3]. The knee was represented as a 1 DoF joint with anteroposterior translation occurring as a function of knee flexion. The model was first scaled to match each subject, then an optimization algorithm was used to generate muscle forces to drive the model. The mass and inertial properties of the shank were modified when modelling the residual limb. The ankle muscles were removed and the ankle joint was actuated with a torque actuator to simulate the prosthesis. Finally, the KCF was determined using the vector sum of the knee reaction force and the sum of forces from muscles crossing the knee joint [4].

KCF was quantified by the peak value and expressed on the tibia in the anteroposterior (AP), axial and mediolateral (ML) directions. Muscle co-contraction was calculated as the ratio of muscle forces from hamstrings and quadriceps when the peak axial KCF occurred. Dunnett's t-test was used to compare the limb effect at each vacuum level with the non-amputees. Two-factor RANOVA was used to determine the effects of vacuum level and limb within the amputees. Post-hoc tests were used for multiple comparisons. Significance was set at $p<0.05$.

RESULTS

The vacuum level effect ($p=0.001$) was significant for the peak axial KCF (Table 1), which was smallest at 15 inHg for both intact and residual limbs. Amputees showed a larger peak axial

KCF on the intact limb at most vacuum levels except for 15 inHg compared to non-amputees. The limb effect was significant on peak AP KCF and the value was larger on the intact limb than residual limb and non-amputees. The peak ML KCFs were larger on the intact limb at the extreme vacuum levels (0 and 20 inHg) compared with non-amputees ($p=0.018$ and 0.016).

The limb effect was significant for muscle co-contraction at the peak axial KCF ($p=0.011$) and the value was smaller on residual limb than intact limb and non-amputees except for no vacuum (Table 2). The vacuum level effects were significant for hamstring force ($p=0.003$) and amputees showed a larger hamstring force at 20 inHg in both residual limb ($p=0.025$) and intact limb ($p=0.021$) than non-amputees.

Table 1: Mean (SD) peak KCF in body weight. I-Intact limb, R-Residual limb

Vacuum Level	0 in Hg		5 in Hg		10 in Hg		15 in Hg		20 in Hg		Control
	I	R	I	R	I	R	I	R	I	R	
Axial ‡	4.40* (0.29)	3.78 (0.46)	3.96* (0.25)	3.54 (0.39)	3.94* (0.27)	3.35 (0.47)	3.75 (0.27)	3.19 (0.47)	4.40* (0.56)	4.04 (0.53)	3.52 (0.38)
AP †	1.77* (0.54)	0.88 (0.44)	1.79* (0.46)	0.75 (0.51)	1.80* (0.44)	0.95 (0.44)	1.68* (0.33)	0.89 (0.33)	1.77* (0.38)	1.05 (0.57)	1.16 (0.46)
ML	0.27* (0.03)	0.21 (0.06)	0.27 (0.04)	0.22 (0.08)	0.26 (0.04)	0.22 (0.07)	0.27 (0.03)	0.21 (0.09)	0.26* (0.03)	0.21 (0.09)	0.23 (0.02)

Table 2: Mean (SD) Muscle co-contraction and muscle force in BW at peak axial KCF.

Vacuum Level	0 in Hg		5 in Hg		10 in Hg		15 in Hg		20 in Hg		Control
	I	R	I	R	I	R	I	R	I	R	
Muscle Co-contraction†	1.88 (1.02)	1.58* (1.18)	2.47 (0.89)	0.95* (0.54)	2.41 (1.63)	0.94* (0.56)	2.66 (1.17)	0.96* (0.49)	1.20 (0.94)	0.92* (0.45)	2.93 (1.79)
Quadriceps †	2.07 (0.63)	1.33 (0.22)	2.00 (0.41)	1.11 (0.46)	1.99 (0.51)	1.09 (0.41)	1.64 (0.82)	1.03* (0.40)	1.65 (0.71)	1.31 (0.45)	1.57 (0.44)
Hamstrings‡	1.29 (0.45)	1.13 (0.52)	0.90 (0.33)	1.39 (0.55)	1.06 (0.50)	1.30 (0.33)	0.80 (0.42)	1.13 (0.17)	1.69* (0.58)	1.54* (0.39)	0.90 (0.49)

†Significant limb effect in amputee group ($p < 0.05$); ‡Significant vacuum level effect in amputee group ($p < 0.05$); *Significant difference with non-amputee ($p < 0.05$).

DISCUSSION

The intact limb showed a comparable peak axial KCF at 15 inHg with non-amputees and the residual limb had a smaller magnitude at 15 inHg than any other vacuum level. This finding could suggest that a major factor associated with knee osteoarthritis, KCF, can be modified by vacuum level using EVSS. This reduction of KCF was noted especially in the intact limb. The muscle co-contraction at peak axial KCF was similar between intact limb and non-amputees at 5, 10 and 15 inHg, but both were larger than the residual limb. Therefore, a moderate vacuum level of 15 inHg is suggested in the design and fitting of EVSS for unilateral TTA.

REFERENCES

1. Klute, G.K., et al. (2011) Arch Phys Med Rehab, 92(10):1570-1575.
2. Goswami, J., et al. (2003) Prosthetics and Orthotics International, 27(2):107-113.
3. Delp, S.L., et al (1990) IEEE Transactions on Biomedical Engineering, 37(8): 757-767.
4. Delp, S.L., et al (2007) IEEE Transactions on Biomedical Engineering, 54(11):1940-1950.

DISCLOSURE STATEMENT

There are no conflicts of interest to disclose.

KINEMATICS COMPARISON OF EARLY KNEE VERSUS TRADITIONAL KNEE PROSTHETIC PRESCRIPTION IN YOUNG CHILDREN WITH AMPUTATION

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INTRODUCTION

A well-established clinical standard dictates that young children with limb loss at or above the knee joint do not receive a prosthetic knee that flexes and extends until they are capable of independent standing and walking. This protocol was established for two main reasons. First, a general assumption was made that in the development of upright balance and walking, stability is preferred over mobility. Second, past prosthetic knee joints were not manufactured to be of appropriate size for infants and toddlers [1]. Recently, pediatric flexing knee joints have been developed, so the second assumption is no longer valid. On the other hand, the first assumption is still widely believed to be true, since the control of a passive prosthetic knee joint requires balance, coordination, and selective firing of residual limb hip flexors and extensors and other proximal muscles at appropriate phases of the gait cycle. Recent research has shown that young children are actually capable of crawling and walking with an articulating prosthetic knee and that several gait parameters are improved with an articulating knee versus a locked knee [2,3]. While promising, there is a lack of formal comparison to children in the traditional protocol. The purpose of this study was to compare the biomechanical impact of the two protocols in children in the Early Knee protocol to children in the traditional protocol.

CLINICAL SIGNIFICANCE

The outcomes of this research could change the timing of when young children receive a working prosthetic knee. The use of a knee at an early age, even before the development of standing and walking, can facilitate the development of associated motor and cognitive milestones at more typical ages.

METHODS

Gait analysis was performed on two groups of children (N=4 in each group) five years of age or younger with unilateral lower limb loss. The Early Knee protocol (EK) children were assessed at Georgia State University, and children with unilateral lower limb loss in the traditional protocol, with no articulating knee (TR), were assessed at Shriners Hospitals for Children in Shreveport, Louisiana. The protocol was approved by human subjects review boards at each institution, and informed parental permissions were obtained. Kinematic data were collected at both laboratories at 100 Hz using a Vicon motion analysis platform running Nexus and Polygon for data acquisition and analysis (Oxford Metrics, Oxford, UK). The standard Vicon lower body marker set was implemented at each site, and at least ten walking trials were captured at comfortable, self-selected speed.

RESULTS

The study is designed for between-subjects age-matched paired comparisons. At this preliminary point average age, height and body mass for ER group is 51.25 (18.96) months, 15.85 (6.58) kg and 103.0 (17.87) cm and for the TR group 27.25 (9.0) months, 13.55 (3.40) kg and 88.85

(13.40) cm. Although paired comparisons are not yet feasible, mean results do show that the EK group utilized the prosthetic knee for swing phase flexion averaging 66.5 degrees (Figure 1). Bilateral asymmetries are apparent in both groups (Table 1). The EK group on average was slower but took longer steps.

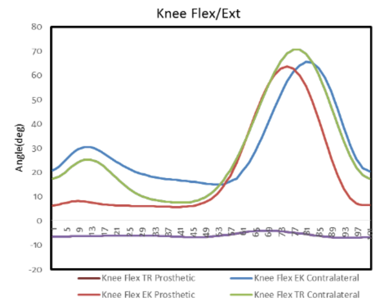

Figure 1: Mean knee flexion

DISCUSSION

This study is the first direct comparison of the early knee protocol to the traditional pediatric prosthetic prescription protocol. It establishes that young children who are provided with a flexing prosthetic knee at an early age are capable of using the knee for flexion. It would therefore be expected that clearance adaptations would be reduced in this group. Some in the EK group did, however, exhibit signs of upward pelvic obliquity during swing phase, a measure of hip-hiking, possibly due to the nature of amputation in that group (proximal femoral focal deficiency). This type of transfemoral above knee prosthesis could need different prosthetic limb adjustments such as socket trim design which might limit the pelvic of the prosthetic side to drop. The children in the locked knee group exhibited a trend toward a greater hip joint frontal plane motion, especially for hip abduction in the swing phase. Swing phase hip abduction is a measure related to circumduction for clearance. Additionally, maximum ankle plantar flexion in the TR group was larger than the ER knee group. This could be a strategy for swing phase toe clearance. Additional subjects will enable like-to-like comparisons based on age and level of amputation.

REFERENCES

1. Wilk, BS. et al. (1999) J Prosth Orth ;11(3):69-74.
2. Geil, MD. et al. (2014) Prosth Orth Int ;38(1):54-61.
3. Geil, MD. et al. (2010). J Prosth Orth ;22(1):21-5.

ACKNOWLEDGMENTS

This study was funded by a grant from the Gerber Foundation.

DISCLOSURE STATEMENT

The authors have no conflicts of interest to disclose.

Table 1: Mean(SD) values for gait parameters for two knee protocol groups

	EK Contralateral	EK Prosthetic	TR Contralateral	TR Prosthetic
Gait parameters	Average (SD)	Average (SD)	Average(SD)	Average(SD)
Walking Speed(m/s)	0.67 (0.31)		0.72(0.40)	
Step length (m)	0.38 (0.14)	0.37(0.15)	0.33(0.11)	0.28(0.12)
Max Pelvic Obliquity	0.04 (4.37)	13.40(6.84)	5.23(8.23)	4.33(2.83)
Hip flex initial contact	49.09 (7.72)	19.83(8.74)	42.51(10.38)	23.21(21.69)
Max Hip extension	-4.69(10.64)	-13.20(8.31)	-7.03(3.07)	-13.64(7.44)
Max hip abduction	-17.20(10.19)	-9.68(8.47)	-18.81(6.50)	-12.95(7.61)
Max knee extension	13.34(7.06)	5.66(7.40)	5.83(3.38)	NA
Max knee flex in swing	72.83(6.40)	66.05(29.55)	74.10(5.85)	NA
Max ankle plantarflexion	-2.37(12.23)	13.00(18.75)	-6.12(4.39)	4.28(1.23)

Poster Session #3: Even Posters

	TITLE	AUTHORS
2	A Comparison of EMG Normalization Techniques in Gait Initiation and Perturbation Studies	Aisha Moore, Pouye Sedighian, Nicholas Gardiner, Vennila Krishnan, Shadnaz Asgari
4	A 13 Year Follow-up of Split Tendon Transfers for Equinovarus Deformity Secondary to Cerebral Palsy	Ann Flanagan, Peter A Smith, Joseph Krzak, Karen Kruger, Adam Graf, Sahar Hassani
6	Calculating the Area of Compressed Plantar Tissue During Gait	Daniel Lidstone, Jessica DeBerardinis, Louise Porcher, Mohamed Trabia, Janet Dufek
8	An Age Comparison of Gait Mechanics While Ascending an Inclined Surface	Jeffrey B. Casebolt, Kunal Singhal
10	Determining Gait Symmetry Using Pressure-measuring Insoles	Jessica DeBerardinis, Daniel Lidstone, Janet Dufek, Mohamed Trabia
12	The Role of Expectation Bias in Gait Characteristics and Orthosis Perception	Brittany Balsamo, Mark Geil
14	Muscle Strength: A Key Factor in Walking Function in Children with Cerebral Palsy.	Annie Pouliot-Laforte, Audrey Parent, Yosra Cherni, Martin Lemay, Laurent Ballaz
16	Inverse Kinematics Using Portable, Low-cost Sensor Technology	Stephen Glass, Alessandro Napoli, Iyad Obeid, Carole Tucker
18	Categorization of Gait Patterns Based on Ankle and Hip Stiffness in Patients with Hemiparesis Due to Stroke	Yusuke Sekiguchi, Takayuki Muraki, Dai Owaki, Keita Honda, Shin-Ichi Izumi
20	The Intra-Rater and Inter-Rater Reliability of a New Modification of the Ely Test for Individuals with Cerebral Palsy	Lisa Drefus, Siobhan Clarke, Karen Resnik, Emily Dodwell, David Scher, Jayme Burket
22	The Development and Treatment of Adolescent Hip Pain in a Patient with Legg-Calve-Perthes Disease	Kirsten Tulchin-Francis, David Podeszwa, Adriana De La Rocha, Daniel Sucato
24	Differences in Squatting Biomechanics in Individuals with Unilateral and Bilateral Adolescent Hip Dysplasia	Alicia Y. Kokoszka, Wilshaw R. Stevens Jr., David A. Podeszwa, Kirsten Tulchin-Francis
26	Pilot Study on the Effects of Upper-limb Loss and Prosthesis Use on Locomotor Stability	Rebecca Stine, Matthew Major, Suzanne McConn, Steven Gard
28	Walking Sticks Vs. Walker on Spine Kinematics in Adult Scoliosis Patients Pre and Post Surgery	Ram Haddas, Isador Lieberman
30	Case Study: Use of Opensim Musculoskeletal Modeling to Optimize Orthotic Tuning in a Patient with Hemiplegic Cerebral Palsy	Nancy Scullion, Haluk Altiok, Christina Garman
32	Protective Stepping In People With Multiple Sclerosis: Effects Of A Single Bout Of Practice	Daniel Peterson, Kris Kratz, K. Bo Foreman, Lee Dibble

Poster Session #3 (continued)

	TITLE	AUTHORS
34	Impulse and Muscle Activity of the Leading Limb During Backward and Lateral Lunges As Well As the Trailing Limb During Forward Lunge	Kay Cerny, Janet Adams
36	Biofeedback Results in Gait Characteristics Similar to Controls with Decline Walking After Total Knee Arthroplasty	Jesse Christensen, Paul LaStayo, Ryan Mizner, K. Bo Foreman
38	The Change of Thoracic Spine Angle and Lumbar Spine Angle During Speed-Up Running	Michio Tojima, Ayaka Osada, Suguru Torii
40	Comparison of Knee Joint Force and Moment During Short and Middle Turns of Carved Ski Using Wearable Motion Analysis System	Yoon Hyuk Kim, Jun Seok Kim, Purevsuren Tserenchimed, Kyungsoo Kim
42	Comparison Of Linear And Non-linear Approaches To Estimating Scapular Motion	Kristen Nicholson, Tyler Richardson, Elizabeth Rapp, Garry Quinton, Kert Anzilotti, James Richards

A COMPARISON OF EMG NORMALIZATION TECHNIQUES IN GAIT INITIATION AND PERTURBATION STUDIES

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INTRODUCTION

The amplitude of electromyography (EMG) signal is influenced by various confounding factors, including the anatomical and biochemical characteristics of the muscles [1]. Hence, EMG signal normalization (i.e., scaling the signal's amplitude relative to a reference value) is a crucial step for the correct interpretation of the signal [2]. While various normalization techniques have been proposed over the last few decades, there is still a debate on which method is the most robust to the intra-subject and inter-subject variability [2-4]. Several studies have looked at the effect of different normalization techniques during walking [2-3]. But to the best of our knowledge, no study has yet investigated the effect of these techniques on gait initiation or perturbation. This work is aimed at the comparison of two popular normalization techniques (static and dynamic approaches) in the gait initiation and perturbation studies.

CLINICAL SIGNIFICANCE

EMG signal analysis can provide valuable information about the muscle activity. However, to properly quantify EMG data, an appropriate normalization technique is needed for the subject to subject, and/or muscle to muscle comparisons [1,5].

METHODS

Our study cohort consists of 10 healthy right leg-dominant subjects (5 female/5 male, age range of 18-30 years old). The EMG electrodes were attached to the following 10 muscles on the right side of the body: soleus (SOL), erector spinae lateralis (ESL), gluteus medius (GMED), tibialis anterior (TA), biceps femoris (BF), external obliques (EO), vastus medialis (VM), medial gastrocnemius (GM), rectus femoris (RF), rectus abdominis (RA)).

The subjects pushed and pulled a fixed bar with maximal effort for 5 seconds to collect the static peak contraction value for the muscles that opposed the action [4].

The subjects then stood on an AMTI force plate and were perturbed with a pendulum at the shoulder level in both front and back directions with both eyes open and closed (4 experiments at a random sequence) for a total of 15 trials per experiment. The subjects also stood on the force platform and commenced gait initiation at a self-selected speed with and without 15% body weight added around the pelvis (2 experiments) for a total of 15 trials per experiment. An accelerometer was placed on the left clavicle during perturbation and on the right knee during gait initiation to calculate the moment at which the subject began to move (T0). For each experiment, the collected EMG data within a two-second window (centered at T0) was used to obtain the normalization values using the dynamic approach.

The EMG signal was rectified and bandpass filtered (Butterworth filter with transition band of 10-50Hz), and integrated by applying a moving window of 20 milliseconds (IEMG). For the dynamic normalization of perturbation and gait initiation data, the peak IEMG values were calculated over all trials of each experiment for each muscle [6]. The peak values of IEMG during pull/push experiment were employed for static normalization. Finally, a t-test was performed to compare the normalization values of dynamic and static approaches.

RESULTS

Figure 1 demonstrates the boxplots of the normalization values, and Table 1 lists the P-values of the t-test on whether the dynamic values are significantly higher than the static (P-value<0.05). It was found that for perturbation (and gait initiation), the dynamic approach resulted in higher normalization values than the static approach for all the muscles, except the GM and BF muscles in perturbation (and ESL muscle in gait initiation).

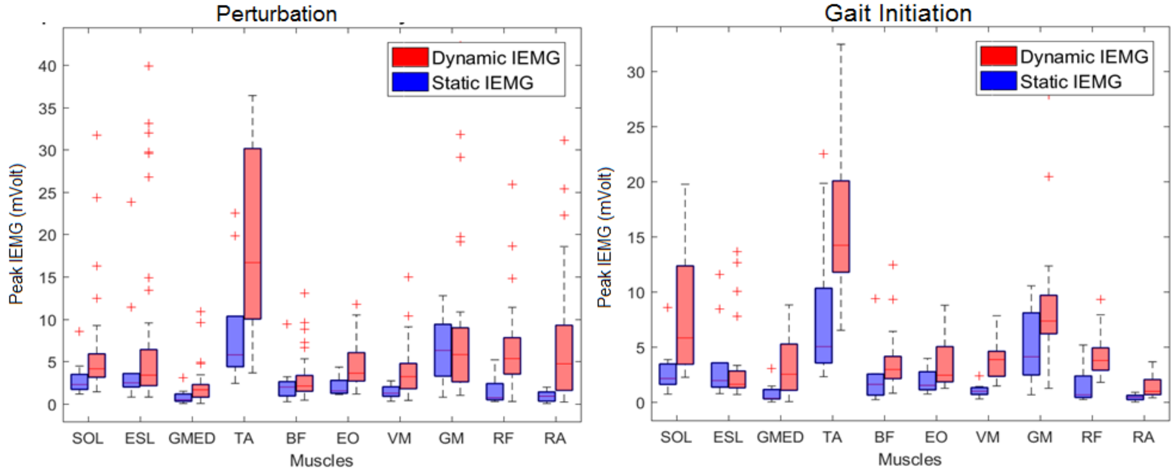

Figure 1: Boxplots of the normalization values obtained from static and dynamic approaches

Table 1: P-values of the t-tests for Perturbation (Pert) and Gait Initiation (GI)

Muscle	SOL	ESL	GMED	TA	BF	EO	VM	GM	RF	RA
Pert	.0013	.0008	.0012	.0000	.1769	.0000	.0000	.1052	.0000	.0000
G	.0001	.5599	.0004	.0000	.0209	.0013	.0000	.0088	.0000	.0002

DISCUSSION

Our results indicate that the dynamic and static approaches lead to significantly different normalization values in both perturbation and gait initiation. Given that the majority of the muscles had a higher normalization value with the dynamic approach, we would recommend using this approach for normalization in gait initiation and perturbation studies.

REFERENCES

1. Danoff, Jerome V. "EMG normalization." *Physical therapy* 66.2 (1986): 270-272.
2. Mantoan, Alice, et al. "Estimation of muscle forces based on a multi-DOF EMG-driven neuromusculoskeletal modeling approach: Impact of different EMG normalization strategies." *Gait & Posture* 42 (2015): S2.
3. Sinclair, J., et al. "The efficacy of EMG MVC normalization techniques for running analyses." *Journal of Biomechanics* 45 (2012): S623.
4. Piscitelli, Daniele, et al. "Anticipatory postural adjustments and anticipatory synergy adjustments: preparing to a postural perturbation with predictable and unpredictable direction." *Experimental Brain Research* (2016): 1-18.
5. Sinclair, Jonathan et al. "The Reliability of Electromyographic Normalization Methods for Cycling Analyses." Journal of Human Kinetics 46 (2015): 19–27.
6. Sartori, Massimo, et al. "EMG-driven forward-dynamic estimation of muscle force and joint moment about multiple degrees of freedom in the human lower extremity." *PloS one* 7.12 (2012): e52618.

DISCLOSURE STATEMENT

There are no conflicts of interest to disclose.

A 13 YEAR FOLLOW-UP OF SPLIT TENDON TRANSFERS FOR EQUINOVARUS DEFORMITY SECONDARY TO CEREBRAL PALSY

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INTRODUCTION

Equinovarus (EV) foot deformity is commonly seen in ambulatory patients with cerebral palsy (CP) and interferes with stability in stance, pre-positioning the foot in terminal stance, and clearance in swing during gait. Soft tissue surgeries such as a split anterior (SPLATT) and/or posterior (SPOTT) tibial tendon transfer are indicated to balance coronal and transverse plane torque throughout the foot.

CLINICAL SIGNIFICANCE

Limited long-term reports on outcomes had variable results. The purpose of this study was to examine the long-term outcomes of these procedures.

METHODS

Ten participants (average age 22y 10m; 4M, 6F; 5 hemiplegia, 4 diplegia, and 1 triplegia involvement; Gross Motor Functional Classification System (GMFCS) I n=7 and GMFCS II n=3) who had EV corrective surgery at an average of 9y 2m, were seen for a follow up visit in the Motion Analysis Lab an average of 13y 6m post-surgery. Outcomes included multi-segment foot and ankle gait kinematics using the Milwaukee Foot Model, visual analog pain assessment, ambulatory ability, static foot alignment based on Kling criteria, and hindfoot score based on Chang. Surgeries included four SPLATT and nine SPOTT procedures. Five participants had follow-up surgery.

RESULTS

Gait kinematics demonstrated that the Equinovarus Group presented with mild residual equinus (hindfoot plantar flexion and decreased forefoot plantar flexion), hindfoot valgus, and forefoot varus (Figure 1). Residual/recurrent transverse plane foot deformity was not appreciated. Average pain score in the foot was 2.5 out of 10. Nine participants ambulated community distances and one was a short community ambulator. All wore regular shoes, and five participants had callouses at their great toe. Using the Kling foot positioning criteria, there were six excellent and four good results. In contrast to hindfoot kinematics, the Chang hindfoot grading criteria results showed seven neutral, one mild varus and two mild valgus feet after correction.

DISCUSSION

Overall, surgical results were maintained 13 year post-operatively. Corrections were maintained in most subjects' hindfoot and forefoot coronal plane alignment and overall foot gait kinematics; however mild residual equinus did persist. Pain and callouses were minimal and all subjects now fit into regular shoes and attend school/work full time. Findings support the use of split tendon transfers to treat equinovarus foot deformities. The results also illustrate the importance of objective gait kinematics as observational-based tools did not detect residual, recurrent hindfoot valgus.

ACKNOWLEDGEMENTS: We would like to acknowledge the Helen Kay Charitable Trust grant for funding this study.

DISCLOSURE STATEMENT: None of the authors have conflicts of interest to disclose.

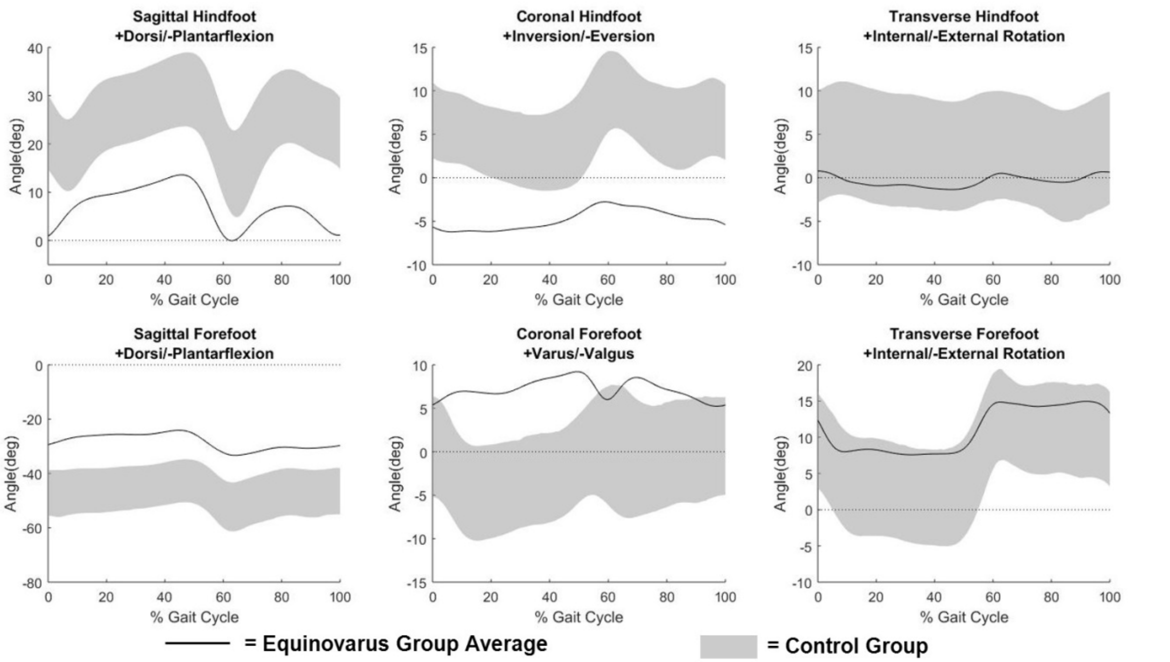

Figure 1. Multi-segment kinematics of the Control Group (Grey band: Mean +/- 1 SD) and the Equinovarus Group (Black line: Mean)

CALCULATING THE AREA OF COMPRESSED PLANTAR TISSUE DURING GAIT

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INTRODUCTION

Previous investigations have highlighted the importance of measuring plantar surface dynamically for the identification of foot deformities and pathology [1, 2], gait parameters [3], and plantar kinetics [4]. Researchers have claimed obtaining plantar surface area dynamically, in comparison to static measurements, is more functional and may provide more valuable information related to foot pathology [1, 2]. A pedography technique that can quantify the compression area of the foot during dynamic tasks may therefore provide valuable information on foot pathology and pathological gait. The current investigation proposes a digital imaging technique to measure plantar surface area during the contact phase of gait through direct imaging of the compressed plantar surface in combination with an algorithm that monitors change of coloration of this surface.

CLINICAL SIGNIFICANCE

The development of a low-cost technique that can accurately image the plantar surface area dynamically can aid clinicians in identifying foot deformities and gait pathologies during gait.

METHODS

The current study was approved by the affiliated Institutional Review Board. The participants (n = 3; 2 male, 1 female) in the current investigation performed 10 barefoot walking trials across a 3 m long transparent walkway. A high definition camera (Nikon 1 J4, Nikon), centered below the middle of the walkway, imaged plantar surface contact at 60 Hz while the subject was walking at a self-selected speed. The camera position was standardized to be centrally located 0.6 m below the walkway. The camera was placed in a custom 3D printed casing, to ensure consistency of the experiments. The camera zoom was fixed for all walking trials. The walkway was illuminated with standardized lighting settings using LEDs located above and underneath the transparent walkway. The camera was calibrated prior to the data collection using the known size of squares on a checkerboard. The subject initiated each walking trial stepping with their right foot. The starting position of the subject was standardized so that the subsequent right foot contact occurred at the center of the transparent walkway.

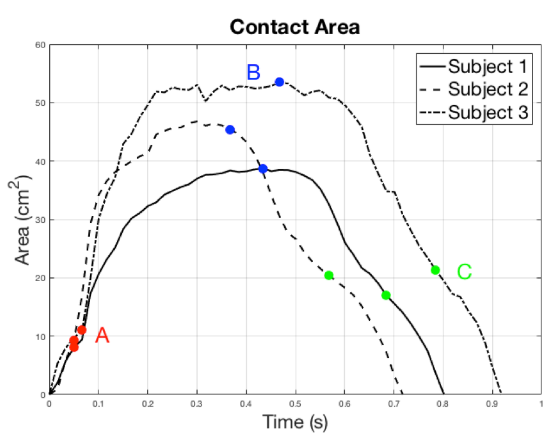

Figure 1: Contact areas during step for heel contact (A – red), full foot contact (B – blue), and metatarsal contact (C – green) generated from the coloration-tracking algorithm for Subject 1, 2, and 3 with US men's shoe sizes 6.5, 8, and 10.

Following data collection, the videos were processed in Matlab (Mathsoft) with custom code that used several Matlab toolboxes, including Image Processing. The RGB (red, green, blue) values of the pixels within the plantar contact area were inputs into a custom developed coloration-tracking algorithm. Pixels with RGB values within a specified multiple of the standard deviation were automatically identified as being in contact with the walkway platform. These areas were identified as white in the processed image. All other pixels were identified as black in the processed image. This was performed for each frame during the contact period. The pixels within the identified contact area were summed and converted to cm^2 for each image (Figure 1).

DEMONSTRATION

Figure 2 shows a comparison between the actual and processed images using the automated coloration-tracking algorithm for three phases of the contact period.

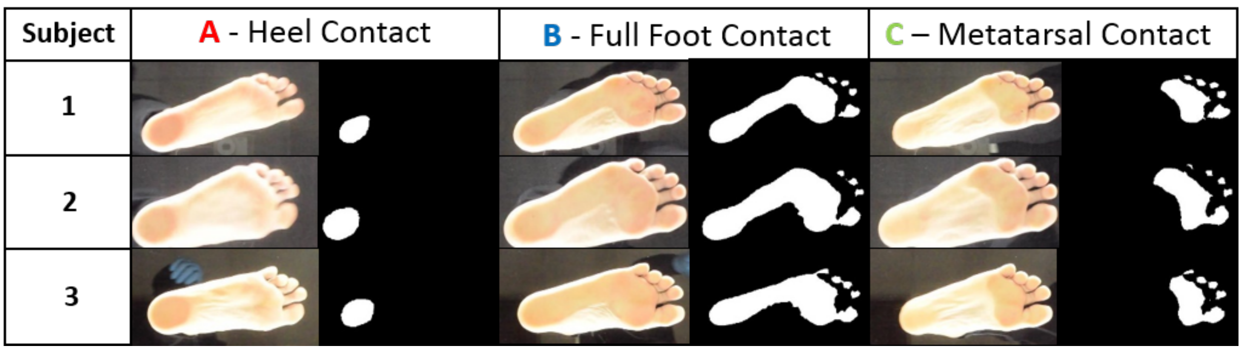

Figure 2: Contact area images from video taken underneath the transparent walkway and processed with black and white images generated from the coloration-tracking algorithm.

SUMMARY

The current investigation has demonstrated the effectiveness of a low-cost, robust coloration-tracking algorithm to identify the plantar surface contact area during gait. We produced plantar contact area-time curves for three participants, each with different foot size. At full foot contact (B – blue), clear differences between the subjects were observed (Figure 1). The smallest area at B corresponded to the subject with the smallest foot size and the largest contact area corresponded to the subject with the largest foot size. Qualitative analysis of the actual vs. processed images clearly demonstrates the effectiveness of this technique to replicate the dynamic footprint of each participant. This technique may assist clinicians to quantitatively identify pathological gait and foot deformities during gait.

REFERENCES

1. McPoil, T. et al. (2009) Journal of Foot and Ankle Research, 2(1):28
2. Mathieson, I. et al. (1999) The Foot, 9:145-149
3. Wilkinson, M.J. & Raspovic (1995) The Foot, 5(2):84-90
4. Fascione, J. et al. (2014) Journal of American Podiatric Medical Association, 104:125-133

DISCLOSURE STATEMENT

There are no conflicts of interest to disclose.

AN AGE COMPARISON OF GAIT MECHANICS WHILE ASCENDING AN INCLINED SURFACE

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INTRODUCTION

An estimated 30-50% of elderly fall each year [4,5]. Mechanically, there is an increased demand on the body when walking uphill. Because ramps are preferred among the elderly when both stairs and ramps are available do to the high risk of injury if a fall should occur on stairs [1,4,5] citing reduced falls, less muscular effort, and gait patterns similar to level walking [2]. Additionally, gait mechanics changes significantly with age with older adults showing increased trend towards maintaining stability compromising on mobility [3]. This trend along with increased demand placed on the body due to increased demand necessary to raise the body's center of mass by altering the joint positions and kinetics. As a result fall risk among the elderly can be compromised. However, little research has addressed elderly gait and safety when ascending a ramp. Therefore, the purpose of this study was to determine age and slope associated change in gait mechanics which may increase the fall risk for older adults.

CLINICAL SIGNIFICANCE

To determine the age associated changes in human gait which may contribute to increased fall risk when descending a ramp at 5˚ and 10˚ in comparison to level walking.

METHODS

Two groups of participants (35 young = 18–30; 29 elderly = 60+ years of age) granted informed consent prior to participation in the study. An 8.5 meter walkway with four AMTI OR5-6 force plates embedded within the path was constructed to simulate three walking conditions: level, and descending 5˚ and 10˚. The body model consisted of an eight segment rigid body consisting of trunk, pelvis and lower extremity segments. Markers were tracked using a 10 camera Vicon system (Vicon, Oxford, UK) and analyzed in Kwon 3DXP (Visol Inc, Seoul, South Korea). The reconstructed marker coordinates were filtered using a Butterworth low-pass filter (4th order zero lag filter) with a 6 Hz cut-off frequency.

A 2 x 3 mixed factorial design was used for age and walking conditions (0°, 5°, and 10°). The dependent variables were static coefficient of friction during stance phase, gait velocity, stride length, GSR, sole inclination, toe clearance, lateral sway ratio, and ankle, knee, and hip joint powers. A repeated measures MANOVA was used for statistical comparison followed by univariate test and a Bonferroni correction in case of significant difference. Alpha level (α) was set *a priori* at 0.05.

RESULTS

Table 1: Variables: Gait Velocity, Step Length, Gait Stability Ratio (GSR), Sole Inclination, Ankle Concentric Power (ACP), Knee Concentric Power (KCP).

		Gait Velocity (m/s)			Step Length (m)			GSR (steps/m)		
		0˚	5˚	10˚	0˚	5˚	10˚	0˚	5˚	10˚
Elderly	Mean	1.41	1.38	1.30*#	0.734*	0.741*	0.700*%	1.38*	1.35*	1.35*
(n=29)	*SD*	*0.18*	*0.15*	*0.19*	*0.08*	*0.07*	*0.08*	*0.13*	*0.13*	*0.13*
Young	Mean	1.47	1.45	1.44	0.796	0.811	0.810	1.27	1.24	1.23
(n=35)	*SD*	*0.16*	*0.17*	*0.15*	*0.05*	*0.06*	*0.07*	*0.13*	*0.13*	*0.14*
		Sole Inclination (α)			ACP (W/kg)			KCP (W/kg)		
		0˚	5˚	10˚	0˚	5˚	10˚	0˚	5˚	10˚
Elderly	Mean	22.53*	18.59*$	12.95*#%	3.21*	3.79*$	4.17*#	0.67	1.13*$	1.38*#%
(n=29)	*SD*	*3.27*	*4.85*	*4.11*	*0.71*	*0.87*	*0.80*	*0.31*	*0.50*	*0.45*
Young	Mean	27.67	23.37	18.77	3.83	4.41	5.16	0.70	1.00	1.64
(n=35)	*SD*	*3.35*	*4.48*	*4.90*	*0.86*	*0.99*	*0.90*	*0.24*	*0.41*	*0.61*

*indicates significant age differences α ≤ 0.05
#indicates significant slope difference α ≤ 0.05: 0˚ to 10˚
$indicates significant slope differences α ≤ 0.05: 0˚ to 5˚
%indicates significant slope differences α ≤ 0.05: 5˚ to 10˚

DISCUSSION

The current study determined elderly are at a higher risk of falling when compared to young individuals while ascending ramps based on the variables investigated. While walking upslope the concentric joint powers were similar for ankle and knee for elderly indicating an concentric control strategy. The elderly showed similar gait speed for level walking; however, at 10˚ upslope there was a significant decrease. The decrease in gait speed with increasing slope was a direct result of a significant decrease or availability of ankle concentric power to forward progression in order to overcome the rate of gravity. Knee concentric power follows a similar trend. As a result, elderly gait velocity and GSR are significantly decreased and increased, respectively at 10˚ down slope. The increase in elderly GSR and decreased sole inclination in comparison indicates a need to increase gait stability by taking more steps per unit of distance and walking with the bottom of their feet closer to the surface. In conclusion, elderly gait is altered due to a decreased ability to produce concentric ankle and knee power resulting in decreased gait velocity and step length, which has been shown to increase chances of a fall.

REFERENCES

1. Wolfinbarger & Shehab (2000). *Proceedings of the IEA 2000/HFES 2000 Congress, 4*, 76-79.
2. Jackson & Cohen (1995). *Journal of Safety Research, 26*, 151-159.
3. Cromwell, R. L., Newton, R. A. (2004). *Journal of Aging and Physical Activity*, 11, 90-100.
6. Hausdorff et al. (2001). *Archives of Physical Medicine and Rehabilitation, 82*(8), 1050-1056.
7. Tinetti et al. (1988). *New England Journal of Medicine, 319*(26), 1701–1707.

DETERMINING GAIT SYMMETRY USING PRESSURE-MEASURING INSOLES

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INTRODUCTION

Gait has been extensively researched to determine 'normal' symmetry as a symptom of pathology [1,2] and as an assessment of a rehabilitative treatment [3]. Pressure-measuring insoles have been used for analysis of gait symmetry [4]; however, this analysis was limited. Participants were wearing shoes, which might have influenced the measurements [5]. Also, there was no comparison to another valid instrument, such as a force platform, to determine if the measured asymmetry was valid.

The purpose of this experiment was to determine the relationship of the impulse symmetry index as quantified by pressure-measuring insoles (Medilogic) in comparison to similar measures obtained from a force platform (Kistler). Analysis included a comparison between the symmetry measures of each instrument and an identification of asymmetric gait patterns of the participants.

CLINICAL SIGNIFICANCE

If the insoles are capable of accurate measurement of gait symmetry, they will be deemed a reliable clinical and rehabilitation tool for clinical gait assessment.

METHODS

A total of 39 healthy, ambulatory adults (14 men, 25 women, 23.5 ± 3.24 yrs., 66.7 ± 17.5 kg, 1.64 ± 0.09 m) gave institutionally approved written consent to participate. The participants were fitted with pairs of pressure-measuring insoles (*Ins*), which were placed next to the skin inside thin socks provided by the researchers. This simulated a barefoot walking scenario. The participants were then asked to perform the following series of tasks, repeated for three trials:

1) Sit on the chair and lift their feet off the floor (3-5 cm) for 5 seconds, and
2) Stand and walk 5 meters over two consecutively mounted force platforms (*FP*), placing one foot on each platform (Figure 1).

Figure 1: Exemplar subject stepping on each individual force platform while wearing the insoles.

Both insole and force platform data were filtered at 1/8 of each instruments sampling frequency, 60 Hz and 1000 Hz, respectively. The filtered data for both insoles and force platforms were normalized by subtracting the bias, which was defined as the non-zero values in the unweighted conditions. The filtered and normalized insole sensor data were added. The insole data were calibrated according to the manufacturer suggested scaling factor of 255 bits equals to 64 N/m^2.

The force data from each instrument were non-dimensionalized with respect to each participant's body weight. Following [6], the symmetry index (*SI*) was calculated based on the impulse of the force-time curves of each instrument as follows:

$$SI_i = 100 * \frac{\left(Imp_{left_i} - Imp_{right_i}\right)}{0.5\left(Imp_{left_i} + Imp_{right_i}\right)} \quad (1)$$

where *i* is *FP* or *Ins.* The average and standard deviation of the symmetry index was calculated from the three trials completed by each participant. Participants who had average symmetry indices of these trials greater than ±4%, [6] were deemed to be asymmetrical and were identified as such. The asymmetric participants identified by each instrument were compared to determine the capability of the insoles to identify the same asymmetric participants as the force platform.

RESULTS

The results given in Table 1 show that while the average symmetry indices for both instruments were similar, the insoles exhibited larger standard deviation values. Table 1 shows a difference in the number of participants identified as asymmetric with the insoles identifying more participants as such. Both instruments identified eleven participants as asymmetric.

DISCUSSION

The results suggest that while pressure-measuring insoles can measure gait impulse reasonably well when compared to the force platform, they produce a very high standard deviation. This is reflected in overestimating the number of participants with gait asymmetry. The difference between the two instruments may be reduced if a more accurate characterization of the insoles is conducted to account for factors such as the insole hysteresis, which is not considered by the current calibration factor [7].

Table 1: The average symmetry indices of all participants and the number of participants identified as asymmetric

Symmetry indices:	**FP Imp (%)**	**Ins Imp (%)**
Mean ± SD:	0.68 ± 3.24	-2.25 ± 13.79
No. of Asymmetric Participants	13	29

REFERENCES

1. Yogev, G. et al. (2007) Experimental Brain Research, 177:336-346.
2. Patterson, K. et al. (2008) Archives of Physical Medicine and Rehabilitation, 89.
3. Archer, K. et al. (2006) Journal of the American Physical Therapy Association, 86:1630-1640.
4. VanZant, S.R. et al. (2001) Journal of the American Podiatric Medical Association, 91(7):337-342.
5. Bates, B.T. et al. (1983) Journal of Biomechanics, 191:181-191.
6. Herzog, W. et al. (1989) Medicine and Science in Sports and Exercise 21(1):110-114.
7. Woodburn, J. et al. (1996) Clinical Biomechanics, 11(5):301-304

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DISCLOSURE STATEMENT

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THE ROLE OF EXPECTATION BIAS IN GAIT CHARACTERISTICS AND ORTHOSIS PERCEPTION

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INTRODUCTION

Biomechanical studies often rely on both laboratory-based objective outcomes and self-reported subjective outcomes. The psychological phenomena of Expectation Bias and Confirmation Bias have been shown to affect user outcomes and behavior, most noticeably when a placebo is shown to produce similar improvements as a treatment. A large body of literature has determined that expectations often lead individuals to actually experience what they expect to experience [1].

In prosthetics and orthotics, significant research effort has been devoted to determining the relative effectiveness of new, advanced-technology components. Because these devices, such as microprocessor-controlled prosthetic knees, are considerably costlier than conventional devices, it is important to establish their efficacy. An unanswered question remains: do users prefer these devices simply because they expect them to perform better. And, moreover, do users actually achieve better objective outcomes in these devices due solely to their expectations?

CLINICAL SIGNIFICANCE

This study shows the role of user perceptions in user-reported outcomes and functional performance, and informs the need for blinding and caution when relying on subjective data.

METHODS

Following informed consent, healthy young adults (N=18) aged 18-26 were presented with two identical off-the-shelf knee braces, but were led to believe that one of the braces, which had been fashioned with a circuit, LED, and USB port, was a computerized brace capable of dynamically adjusting its joint stiffness. Participants were surveyed before seeing the braces about which one they thought would perform better. Then, participants completed non-blinded walking trials with each brace in randomized order. Finally, they were surveyed again about which brace they thought actually performed better, and which brace they would prefer.

RESULTS

Prior to the walking trials, and based only on a description of the braces and a fake "manufacturer's flyer" describing the benefits of the "computerized" brace, 61% of the subjects expressed a strong preference for the "computerized" brace, while 22% preferred the standard brace and 17% expressed no preference. Among the factors assessed, strong preference was shown for the standard brace for Cost, and for the "computerized" brace for Joint Stabilization and Function in Sports.

Following the walking trials, the survey was repeated. This time, 83% of the subjects expressed a strong preference for the "computerized brace", an increase of 22%. The same factors were preferred, but this time strong preference was also shown for the "computerized" brace for Comfort.

Individual results (Figure 1) showed that some participants (5 of 18) confirmed their own expectations after the walking trials, and many (10 of 18) actually increased their overall preference for the “computerized” brace, even though the two braces were identical.

Kinematic and kinetic outcomes revealed no differences in actual walking between the two braces (Table 1). Average walking speed was identical, and average stance and swing phase knee flexion on both the braced leg and the contralateral leg did not exceed 2 degrees.

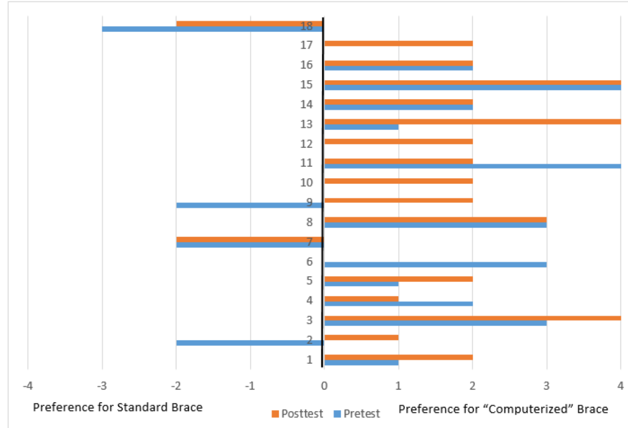

Figure 1: Comparison of Pre-trial and Post-trial questionnaire responses for the “Overall Preference” factor based on a 9-point Likert scale

DISCUSSION

Users showed expectation bias prior to the walking trials. That expectation affected their self-reported feedback following the trials, indicating the presence of confirmation bias. Simply put, the majority of the users preferred one functionally identical brace over another just because they expected it to be better.

By contrast, actual walking patterns (as measured objectively in the biomechanics lab) were unchanged. This result must be considered within the limitations of the study. A healthy population was tested, and since they had no need to improve their walking patterns, expectation might have had less of an impact on those actual walking patterns, even though it had an impact on participants’ perceptions of those patterns. It is possible that participants with impaired gait could reveal actual differences in gait outcomes. Future work could also consider prosthetic components in individuals with limb loss.

REFERENCES

1. Handley, IM, et al. (2013) Int J Beh Med, 20(1):52-8.

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Table 1: Mean kinematic and kinetic laboratory outcomes

Variable	*“Computerized” brace*	*Standard brace*	*p*
Walking speed (m/s)	1.19	1.19	0.59
Stride length, braced side (m)	1.26	1.26	0.95
Stride length, opposite side (m)	1.26	1.26	0.96
Mean peak stance phase knee flexion, braced side (degrees)	43.8	42.1	0.54
Mean peak stance phase knee flexion, opposite side (degrees)	41.8	41.3	0.78
Mean peak swing phase knee flexion, braced side (degrees)	59.5	57.9	0.59
Mean peak swing phase knee flexion, opposite side (degrees)	62.4	62.2	0.95
Mean peak vertical ground reaction force, braced side, peak one (N/kg)	105.1	106.0	0.94
Mean peak vertical ground reaction force, opposite side, peak one (N/kg)	106.9	106.0	0.97
Mean peak vertical ground reaction force, braced side, peak two (N/kg)	110.4	109.4	0.97
Mean peak vertical ground reaction force, opposite side, peak two (N/kg)	110.4	110.6	0.94

MUSCLE STRENGTH: A KEY FACTOR IN WALKING FUNCTION IN CHILDREN WITH CEREBRAL PALSY.

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INTRODUCTION

Cerebral palsy (CP) is associated with muscle weakness and gait impairments. In daily life, the community integration of children with CP depends on their ability to walk independently, but also (1) to walk continuously to reach the surrounding, (2) to have an efficient gait to delay onset of fatigue, and finally (3) to increase walking speed in some challenging situations (i.e. crossing the street) [1]. It has been previously demonstrated that muscle strength is related to gross motor function and gait [2]. In different studies, a higher strength has been related to a higher gait speed or walking efficiency [3]. Nevertheless, in the same group of children with CP, little information is known about the relation between muscle strength and the above-mentioned specific walking ability, which are known as limiting daily life activities. The main goal of this study was to assess the relationship between lower limb muscle strength and different walking function, including walking speed (comfortable and fast), walking endurance, and walking efficiency in children with spastic bilateral CP.

CLINICAL SIGNIFICANCE

A better understanding of the relationship between muscle strength and walking function describe as continuous walking capacity, fast walking capacity and gait efficiency would be useful to highlight the key role of muscle strength in children with CP.

METHODS

A sample of 33 children and adolescents (14 females; mean age ± SD: 11.9 ± 3.7 years; body mass: 36.0 ± 13.5 kg; height: 143.0 ± 19.7 cm; GMFCS II-III) with spastic diplegic CP were included. All participants were able to walk with or without assistive device (orthosis, n=2; canes, n=2; and walker, n=2). A clinical gait analysis (CGA) was performed including three different walking exercises: (1) discontinuous walking as classically done in CGA, (2) continuous 6-minute walking exercise, and (3) fast walking. The children were asked to walk at their comfortable walking speed during the discontinuous and continuous walking exercise, and at their fastest walking speed without running in fast walking condition. The gait efficiency was calculated using the Energy Expenditure Index (EEI), based on heart rate measurement during the continuous walking exercise [4]. Prior to the gait analysis, the maximal isometric strength of the hip abductors and the hip and knee flexors and extensors was measured using a hand-held dynamometer. All values were normalized by the lever arm and by participant body mass (Nm/kg). A global strength index was calculated as the sum of maximal isometric strength from each muscle group evaluated. Pearson product-moment correlation coefficients (r) were calculated to quantify the relationships among all the measures. Five simple regression models, were fitted using an entry selection procedure to quantify the relationship between walking function and the global strength index.

RESULTS

The correlations between walking speed exercises, EEI and muscular strength are presented in Table 1. The average EEI was 1.06 ± 0.6 beats/m. The average walking speed was 60.5 ± 11.9 m/min, 62.0 ±15.1 m/min and 80.5 ±17.1 m/min for the discontinuous, continuous and fast walking speed exercise (significantly different from other conditions, +40%, p<0.001), respectively. Muscle strength correlates with the EEI for all muscle groups but knee flexor. Knee extensors and hip abductors strength correlated with continuous and fast walking speed. All muscle strength groups correlate with the percentage of difference between fast and comfortable walking speed. The global index of strength explains 30%, 21%, 17,5% and 37% of the variance of the EEI, continuous walking speed, fast walking speed and the percentage of difference between fast and discontinuous walking speed, respectively (all p <0.05).

Table 1: Pearson correlation relationship between EEI, walking speed and muscular strength

		Energy Expenditure Index	Discontinuous walking speed	Fast walking speed	Continuous walking speed	Δ fast/comfortable walking speed
Knee extensors	r	-0.50	0.27	0.43	0.43	0.57
	p	0.004**	0.133	0.028**	0.013**	0.002**
Knee flexors	r	-0.29	0.23	0.36	0.37	0.40
	p	0.101	0.202	0.067	0.036**	0.038**
Hip extensors	r	-0.48	0.09	0.34	0.27	0.55
	p	0.009**	0.632	0.090	0.156	0.004**
Hip flexors	r	-0.49	0.12	0.26	0.31	0.43
	p	0.003**	0.520	0.196	0.064	0.026**
Hip abductors	r	-0.46	0.33	0.43	0.48	0.52
	p	0.008**	0.062	0.026**	0.004**	0.005**

DISCUSSION

These results suggest that higher lower extremity muscle strength is associated with all walking function, including continuous walking exercise, fast walking, and walking efficiency. Muscle strength is associated with a greater difference between comfortable walking speed and fast walking speed. No correlation was found between discontinuous waking, as usually performed in CGA, and muscle strength suggesting that maximal strength is key in more challenging condition, (e.i., fast walking and endurance exercise). The present results suggest that walking ability is related to global lower limb strength, as all the significant correlation coefficients are in the same range. In conclusion, our result suggest that therapies should focus on improving muscle strength to improve most aspects of the walking function.

REFERENCES

1. Rose, J. et al. (1989) Journal of Pediatric Orthopedics, 9(3):276-279.
2. Ferland, C. et al. (2012) Physical & Occupation Therapy in Pediatrics, 32(2):320-332
3. Damiano, D L. et al. (1996) Developmental Medicine & Child Neurology, 38(5):389-396.
4. Rose, J. et al. (1990) Developmental Medicine & Child Neurology, 32(4):333-340

DISCLOSURE STATEMENT

The authors have no conflicts of interest to disclose.

INVERSE KINEMATICS USING PORTABLE, LOW-COST SENSOR TECHNOLOGY

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INTRODUCTION

The Microsoft Kinect 2.0™ is a portable platform that allows for low-cost motion capture in field-based and clinical settings [1]. The sensor performs remarkably well as an out-of-the-box technology, particularly considering its limitations as a single-camera, markerless system. These achievements notwithstanding, data from the Kinect™ are not readily comparable to conventional kinematics. Specifically, the sensor generates joint displacement time histories without rigid body definitions or range of motion constraints. These measurement errors tend to have large effects on any joint angles calculated by the user. It is possible, therefore, that the quality of Kinect™ data could be improved for research purposes through the introduction of modeling constraints commonly applied in inverse kinematics. Our purpose is to demonstrate the performance of Kinect-based inverse kinematics solutions in comparison with similar data acquired using standard laboratory technology.

CLINICAL SIGNIFICANCE

Applying modeling constraints to motion capture data acquired using low-cost sensors may increase the accuracy of clinical measurement techniques. These methods could support diagnostics and clinical decision-making by enabling the collection of higher quality data.

METHODS

Eight participants (4 males/4 females, 25.6 ± 2.5 years, age, 170.8 ± 10.1 cm, 66.2 ± 12.8 kg) performed trials of a series of commonly used clinical movement tests as data were captured using 1) a Qualisys Oqus motion capture system (Q), and 2) a Microsoft Kinect 2.0™ (K). Inverse kinematics solutions were computed in OpenSim 3.3 using marker (Q) or joint center (K) trajectories. Both analyses were conducted using the Vicon Plug-In-Gait model with segment/trajectory relationships (i.e. the “marker set”) modified to match the source of the data. Sagittal plane hip and knee angles were analyzed to compare performance between the two systems. After resampling the raw data from the Kinect sensor (which samples at a variable rate), cross-correlation and RMS error were calculated on the synchronized signals.

DEMONSTRATION

Test	**DS**	**HS**	**FSTS**	**ABL**	**MIP**
Hip cor.	0.89 (0.27)	0.90 (0.06)	0.98 (0.01)	0.87 (0.13)	0.80 (0.11)
Hip err.	22.89 (7.51)	14.82 (5.17)	23.40 (6.65)	22.98 (18.02)	23.25 (9.50)
Knee cor.	0.89 (0.25)	0.89 (0.06)	0.99 (0.01)	0.90 (0.08)	0.86 (0.10)
Knee err.	12.41 (8.64)	11.95 (3.45)	14.70 (18.62)	11.70 (5.16	16.00 (6.14)

OHS = Overhead Squat, HS = Hurdle Stepping, FSTS = 5x Sit-to-Stand, ABL = Alternating Barbell Lunge, MIP = Marching in Place, corr. = correlation (*r*), err. = RMS error (deg.)

Figure 1 shows hip flexion time series derived from both sensors during a representative trial of the Five Times Sit-to-Stand test. Kinect tracking of the sagittal plane hip and knee angles using inverse kinematics performs favorably in comparison with a gold-standard system. We do note, however, that whereas temporal features of the signal are well preserved, the magnitude of angular displacements can be overestimated by the Kinect.

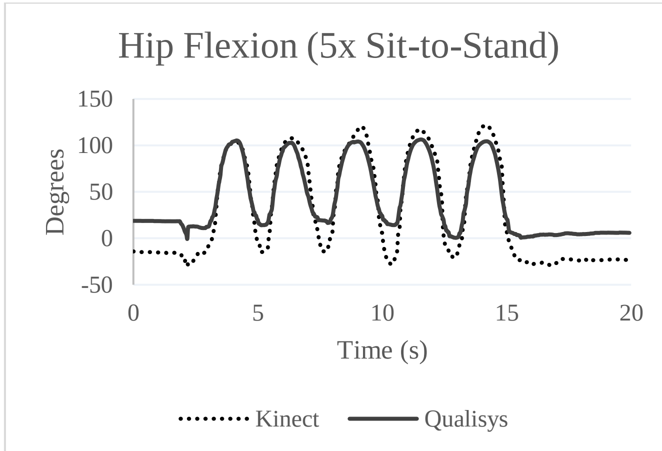

Figure 1

SUMMARY

Our laboratory has recently shown that Kinect 2.0™ data is suitable for instrumenting simple field-expedient clinical tests [2]. With the present work, we expand on our automated scoring algorithm research to improve the quality of this sensor as a robust motion capture tool. These data show that accuracy of certain angular kinematics from low-cost sensors such as the Microsoft Kinect 2.0™ may approach gold-standard criteria with commonly used inverse kinematic modeling techniques. The most substantial benefits likely derive from rigid body definitions and joint-specific range of motion limits, neither of which are applied to Kinect™ data.

Our approach is implemented with openly available software (OpenSim) using a modified version of a familiar kinematic model, the Vicon Plug-In-Gait model. This workflow could greatly benefit clinics and mobile laboratories requiring high quality data without the time and expense typical of multicamera systems. In future work, we will demonstrate the performance of a Kinect-based inverse kinematics analysis in tracking complex, multiplanar movement during a variety of dynamic posture tasks.

REFERENCES

1. Clark, R. A., et al. (2015). Gait Posture 42(2): 210-213.
2. Napoli et al. (2016) EMBC Proceedings, 6090-6093

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DISCLOSURE STATEMENT

The authors have no conflicts of interest to disclose.

CATEGORIZATION OF GAIT PATTERNS BASED ON ANKLE AND HIP STIFFNESS IN PATIENTS WITH HEMIPARESIS DUE TO STROKE

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INTRODUCTION

In general, joint stiffness in the lower limb during gait is an important factor for improvement of gait speed in patients with hemiparesis. Ankle and hip orthoses, which increase ankle joint stiffness (AJS) and hip joint stiffness (HJS), assisted the generation of force in ankle plantar flexion and hip flexion resulting in increased gait speed in patients with hemiparesis [1][2].

In a previous study, gait patterns of patients with hemiparesis were divided into three groups [3]. The categorization of gait pattern was investigated using cluster analysis of kinematic and kinetic data of the lower limb in stroke patients with hemiparesis [3]. However, categorization of gait pattern based on HJS and AJS in patients with hemiparesis remains unclear. Therefore, clinicians may fail to objectively select effective and appropriate orthosis.

The objective of the present study was to classify gait pattern based on AJS and HJS in patients with hemiparesis.

CLINICAL SIGNIFICANCE

The categorization of gait pattern based on AJS and HJS in patients with hemiparesis due to stroke may help in selecting a combination of hip and ankle orthoses.

METHODS

We recruited 50 patients with hemiplegia due to stroke. Inclusion criteria were first-time stroke due to ischemic or hemorrhagic supratentorial lesion, ability to walk without walking prosthetics, and ability to understand instructions by physical therapists. The present study was approved by the local ethics committee of Tohoku University of Medicine, Japan.

Three-dimensional coordinates of 33 reflective markers attached to 12 segments were measured using a three-dimensional motion analysis system with eight cameras operating at 120 Hz. A 12-segment model based on anthropometric data was used as suggested by Dumas [4]. The participants were asked to walk at a self-selected speed without assistance devices. Data analysis was processed using a custom Matlab program. We calculated AJS on the paretic side from the slope of the linear regression between the ankle joint moment and ankle angle during the second rocker interval, which was subdivided into the early (ES) and middle stances (MS). HJS on the paretic side was calculated from the slope of the linear regression between the hip joint moment and hip angle during the period from beginning hip flexor moment to maximum hip extension angle of the stance phase. Moreover, AJS in ES and MS, HJS, and gait speed were subjected to hierarchical cluster analysis. The Ward's linkage method and Squared Euclidean distance measures were the clustering routines applied.

RESULTS

There was a large increase in agglomeration schedule coefficients in the results of hierarchical cluster analysis when four clusters were reduced to three. The participants belonged to one of four subgroups. Subgroup 1 (SG1) included 16 participants, SG2 included 13 participants, SG3 included 17 participants, and SG4 included 4 participants.

There were significant differences ($p < 0.05$) in representative parameters among the four SGs (Table 1).

Table 1: Gait parameters (mean ± SD) in the subgroups (SG)

	SG1	**SG2**	**SG3**	**SG4**
Gait speed (m/s)	0.37 ± 0.13^{b}	0.25 ± 0.12^{d}	0.68 ± 0.14^{bdf}	0.20 ± 0.19^{f}
AJS in ES (Nm/°/kg)	0.013 ± 0.013^{ab}	0.052 ± 0.023^{ade}	0.033 ± 0.018^{bd}	0.016 ± 0.006^{e}
AJS in MS (Nm/°/kg)	0.024 ± 0.013^{ab}	0.053 ± 0.024^{ae}	0.061 ± 0.030^{bf}	-0.008 ± 0.022^{ef}
HJS (Nm/°/kg)	0.024 ± 0.030^{bc}	0.039 ± 0.032^{e}	0.051 ± 0.018^{bf}	0.146 ± 0.044^{cef}
Maximum hip extension angle (°)	-6.1 ± 7.4^{a}	-13.6 ± 6.9^{ad}	-2.0 ± 5.5^{d}	-10.6 ± 8.5

[a] $p < 0.05$ comparing SG1 with SG2. [b] $p < 0.05$ comparing SG1 with SG3.
[c] $p < 0.05$ comparing SG1 with SG4. [d] $p < 0.05$ comparing SG2 with SG3.
[e] $p < 0.05$ comparing SG2 with SG4. [f] $p < 0.05$ comparing SG3 with SG4.

DISCUSSION

We categorized gait pattern based on AJS and HJS in patients with hemiparesis using a hierarchical cluster analysis. The patients were divided into four subgroups that were not similar to the gait patterns used in a previous study based on kinematic and kinetic data [3]. The slow walking groups (SG1, SG2, and SG4) had various characteristics of joint stiffness. This is the first study that demonstrates the type of impairment based on AJS and HJS during gait in patients with hemiparesis.

The lower AJS in MS for SG1 and SG4, which was related to gait speed in a previous study, may lead to the decline in gait speed [5]. Conversely, although AJS in MS for SG2 was not significantly different from that in the fast gait-speed group (SG3), the gait speed for SG2 decreased. This result suggests that there may be other factors that cause a decline in gait speed in SG2. The higher AJS in ES and MS in SG2 could cause a decrease in maximum hip extension angle in stance and result in lower gait speed.

REFERENCES

1. Bregman, DJ. et al. (2012) Gait and Posture, 35(11):148-153.
2. Carda, S. et al. (2012) Phys Ther, 92(5):734-739.
3. Kinsella, S. et al. (2008) Gait and Posture, 27(1):144-151.
4. Dumas, R. et al. (2007) J Biomech, 40(3):543-553.
5. Sekiguchi, Y. et al (2012) Clin Biomech, 27(5):495-499.

DISCLOSURE STATEMENT

We have no conflicts of interest to disclose.

T I r r r dI rr rR N M d E T r I d d C r r P

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INTRODUCTION

The presence of a stiff knee gait pattern due to rectus femoris (RF) muscle spasticity in individuals with cerebral palsy (CP) is well documented in the literature. The Duncan-Ely test is the clinical tool used to assess spasticity of the RF muscle in individuals with CP.[2] This velocity-dependent test is recorded as positive or negative value. A positive Duncan-Ely test is a positive predictor of RF dysfunction during gait. Kay et al.[4] also found the Duncan-Ely test to be a helpful predictor for outcomes in children who are considered for a RF transfer (RFT); improved knee kinematics occurred in the RFT group who had a Duncan-Ely test and no change in the negative group.[4] This study also recorded the knee arc of motion with a goniometer and graded the level of spasticity using by the Modified Ashworth Scale (MAS) point spasticity scale. However, the reliability of the MAS varies in the literature (ICC values 0. to 0. 7)[5,] and is not specific to testing RF spasticity. One study recorded the best inter-rater reliability of the Duncan-Ely test occurs at fast velocity, using the Tardieu scale; but did not test the intra-rater reliability.[7] At our institution, we have used a modification of the Duncan-Ely test quantifying spasticity into a 5 point scale rating where the spasticity occurs through the knee flexion arc of motion. This test was developed and used by Dr. Leon Root over, his nearly 50 year career, treating patients with CP; therefore it is named the Root-Ely test. e have found this to be a more objective way to measure RF spasticity, yet the reliability has not been tested. Therefore, the purpose of this study was to identify the intra-rater and inter-rater reliability of both the Root-Ely and the Duncan-Ely tests. A secondary outcome was to determine if there was a difference among clinicians with various levels of experience.

CLINICAL SIGNIFICANCE

To confirm the reliability of the Root-Ely 5 point scale to measure RF spasticity. This is beneficial to orthopedic surgeons for a more objective scale for predicting ideal candidates for RFT in individuals with CP to improve stiff knee gait.

METHODS:

This was a prospective reliability study recruiting subjects by a sample of convenience. Inclusion criteria included: diagnosis of CP, ambulatory (GMFCS level I, II, III), ages 4 to 21 years, no prior RFT surgery and able to comfortably lie prone on examination table with full knee flexion passive range of motion. The Root-Ely test is a performed by the following steps:

1. The subject lies prone on the table in a relaxed state
2. The examiner places a hand on the pelvis at the sacral region and the other hand on the ankle
. The examiner rapidly flexes the knee through the full knee flexion arc of motion
4. The examiner notes a positive test if the patient simultaneously flexes the ipsilateral hip or resistance is felt by the examiner during the rapid knee flexion
5. The examiner notes where resistance is **FIRST** felt in the arc of knee flexion and/or where the ipsilateral hip flexion is felt.
. The examiner then records the angle of knee flexion, using a universal goniometer, and scores the spasticity using the below 5 point scale in Table 1.

Table 1

Score 4	0-29° of knee flexion and/or significant and sudden pelvic elevation
Score	0-59° of knee flexion and/or moderate pelvic elevation
Score 2	0- 9° of knee flexion and/or mild pelvic elevation
Score 1	_90° of knee flexion and/or minimal pelvic elevation
Score 0 -	No catch or hip rising off table with full knee flexion (negative)

All testers were instructed on the Root-Ely procedure and had the above directions and scale on the test form for each subject. The Root-Ely test was performed during a single session by raters; pediatric physical therapists (PT) and 2 pediatric orthopedic surgeons. Four raters were considered experienced with yrs and 2 raters were considered new clinicians with 5 yrs of working with children with CP. The two primary PT's tested each subject twice for the intra-rater reliability on the same day with __ 0 minute between test 1 and 2. To measure the inter-rater reliability each of the rater tested the subjects one time during a single visit.

Data Analysis: Intra- and inter-rater agreement for the Root-Ely test was assessed with weighted Kappa statistics and 95% confidence intervals (CI). Intra- and inter-rater agreement for the Ely test was assessed with simple Kappa statistics and 95% C. Analyses were performed with SAS 9. (Cary, NC, USA). Based on definitions from Landis and Koch (1977), the measurement of observer agreement for categorical data is: 0. 1-1.00 excellent, 0. 1-0. 0 good, 0.41-0. 0 moderate, 0.21-0.40 fair, and 0.20 poor reliability.

RESULTS

20 subjects (n 40 limbs) with mean age of 10 yrs participated. The Root-Ely intra-rater reliability for rater 1 was 0. 9 with 95% CI (0. 12, 0.979) and rater 2 was 0.7 with 95% CI (0. 19, 0.91). The highest inter-rater reliability between raters was 0. 4 with 95% (0.7 1, 0.9 7) and lowest was 0. 15 with 95% (0.090, 0.5 9). Experienced clinicians inter-rater reliability was excellent to good while new clinicians' inter-rater reliability was moderate to fair. The Duncan-Ely (/-), intra-rater reliability for rater 1 was 0.905 with 95% (0.775, 1.000) and 0. 0 for rater 2 with 95% (0. 29, 0.9 7). The highest inter-rater reliability between raters of the (/-) Duncan-Ely test was 0.790 with 95% (0.597, 0.9 4) and lowest was 0.0 0 (-0.219, 0.2 0). Overall for the Duncan-Ely RF test inter-rater reliability for experienced raters was found to be good to moderate while new clinicians' inter-rater reliability was fair to poor.

DISCUSSION

Previous reports in the literature has validated the use of the /- Duncan-Ely test as a positive predictor for RF spasticity and RFT. [,4] The Duncan-Ely test was found to be the most reliable at a fast knee-flexion velocity with experienced clinicians.[7] This study found good to excellent intra-rater reliability for both the Root-Ely and the Duncan-Ely test. The inter-rater reliability varied similar to MAS reliability studies and was the most reliable among experienced clinicians. Our study adds to the literature as new test that quantifies RF spasticity with more detail than the traditional Duncan-Ely test. Future directions will include testing its ability to predict a stiff-knee gait pattern and the effectiveness of RFT in individuals with CP.

REFERENCES 1. Perry , et al. *Gait analysis: Normal and pathological function.* 2010. 2. Bleck EE. *Orthopaedic management in cerebral palsy* 19 7, . Marks MC et al. *Dev Med Child Neurol*. 200 , 4. Kay RM et al. *J PediatrOrthop*. 2004, 5. Mutla et al. BioMed Central. 200 , . Yam et al. *J Child Neurol*. 200 , 7. Lee S et al, DMCN, 2015

THE DEVELOPMENT AND TREATMENT OF ADOLESCENT HIP PAIN IN A PATIENT WITH LEGG-CALVE-PERTHES DISEASE

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PATIENT HISTORY

The patient is a 15+7 year old male who was first diagnosed with left Legg-Calve-Perthes Disease (LCPD) at age 3yrs (lateral pillar C classification) treated with observation only. The right hip was diagnosed with LCPD at age 7+9years, treated primarily with observation and slight activity modification. Final femoral head deformity was classified a Stulberg 3 on the left and a Stulberg 4 on the right.

CLINICAL DATA

For all visits the patient underwent clinical ROM exam, isokinetic flexion an abduction strength (Biodex System3, 60°/s), and self-reported modified Harris Hip Score (mHHS, normalized to 100pts) as part of a prospective hip pathology research registry.

Visit 1: He was an active young man, playing tennis and marching band, and reported only mild pain with heavy workouts or activity. Pain was not activity limiting and did not warrant intervention. There were slight strength deficits in both flexion and abduction, and his mHHS at 81pts was classified as "Good" (80-89pts).

Visit 2: Approximately one year later, his pain had significantly increased and limited his ability to participate in activities. He reported daily pain, regardless of activity level. His mHHS, now 62pts,decreased by 20 ps and developed a positive anterior impingement sign.

Intervention #1: He underwent a surgical hip dislocation (SHD) with microfracture of the acetabulum, repair of the labrum, femoral osteochondroplasty and relative femoral neck lengthening to address and correct his intra-articular pathology. He was also noted to have mild acetebular dysplasia pre-operatively.

Visit 3: At 6mos post-op, he reported that his pain had improved, however it still limited in his activities and daily routine. Although his mHHS had improved, he had a drastic decrease in hip flexion strength, due to pain during the test.

Visit 4: At 2yrs post SHD, he had continually increasing pain over the last year, He stopped tennis, but continued to do marching band. His mHHS decreased to 62 ("poor"). Due to pain during the passive ROM exam, Biodex strength testing was not conducted.

Intervention #2: Radiographically he had slight under-coverage of the femoral head laterally and anteriorly. He underwent a periacetabular osteotomy (PAO) with psoas lengthening to address his mild acetabular dysplasia and worsening symptoms.

Visit 5: At 2 yrs post-PAO, he was 21 yrs of age and doing extremely well. He reported slight, occasional pain. He works on his feet most of the day and reported no pain with daily activities. He had taken up tennis again.

MOTION DATA

For all visits the patient underwent instrumented gait analysis as part of a prospective hip pathology research registry. Evaluation consisted of overground walking (Vicon Nexus) and 30sec Trendelenberg test (Vicon Nexus/custom MATLAB algorithm).

Table 1: Clinical Evaluation and Self-Reported Outcomes							
Visit	**Age yrs+mos**	**BMI kg/m^2**	**Ant Imp. Sign**	**Flex Strength Nm / % CL †**	**Abd Strength Nm / % CL †**	**mHHS**	**Self-Report Pain**
1	**15+7**	**21.4**	**Slight**	**56.2 / 68%**	**64.8 / 80%**	**81.4**	**Mild**
2	**16+6**	**21.0**	**+**	**70.9 / 85%**	**89.4 / 112%**	**62.7**	**Moderate**
3	**17+1**	**21.4**	**+**	**34.4 / 37%**	**79 / 119%**	**77**	**Mild**
4	**18+4**	**16.9**	**+**	**n/a**	**n/a**	**61.6**	**Moderate**
5	**21+0**	**21.8**	**+**	**64.8 / 75%**	**80.7 / 137%**	**90.2**	**Slight**
† % CL: strength presented as a percentage of the contralateral, asymptomatic side							

Visit 1: He walked with a subtle decrease in max hip extension and knee extension in midstance. Peak pull off power was within normal limits, however he did have a reduction in his coronal plane hip moment, particularly during the second bump in terminal stance which lead to a reduced hip abductor moment in single limb stance

Visit 2: Hip extension was mildly better, however there was a reduction in walking speed and hip flexion power compared to his initial visit. He had a positive Trendelenberg test during a 30s single limb stance, however his hip abductor moment was slight improved.

Visit 3: At 6mos postop, he had reduced walking speed, but improved hip extension. His abduction moment continued to improve as well.

Visit 4: Prior to this PAO, his overall gait pattern was much improved despite continued anterior groin pain.

Visit 5: At final follow-up he had near normal hip kinematics and kinetics.

Table 2: Gait and Motion Analysis Outcomes							
Visit	**Age yrs**	**Speed m/s**	**Min hip flex °**	**Peak Hip Flex Power W/kg**	**Max Hip ABD mom, term st Nm/kg**	**Hip ABD Mom Impulse Nm/kg·s**	**30sec Trendel Test.**
1	15.6	1.2	**8**	1.71	**0.59**	**0.167**	-
2	16.5	1.1	**3**	**0.96**	**0.62**	0.231	+
3	17.1	**1.0**	**0**	**0.88**	0.69	0.252	-
4	18.3	1.1	**3**	1.59	0.76	0.254	-
5	21.0	1.1	**0**	1.37	0.84	0.266	-
Control N=29	17.4 ± 3.2	1.3 ± 0.2	-12 ± 5.6	1.57 ± 0.51	0.79 ± 0.14	0.273 ± 0.053	na

SUMMARY

Femoral head deformity due to LCPD can cause femoroacetabular impingement. The onset of pain in this case example was subtle at first but quickly became worse as the patient went through adolescent growth spurt and continued competitive athletics. With only mild pain at his first visit, he was considered clinically stable. However gait analysis indicated significant reductions in hip extension which also affected the peak hip abduction moment in terminal stance. As his pain increased, his walking speed decreased which may have been a compensation to maintain/improve his gait pattern. Despite continued pain following his SHD he did show improvements in gait. At final follow-up, after staged SHD and PAO, the patient had near normal kinematics and kinetics, significant improvements in pain and function, and excellent clinical outcomes. This longitudinal case demonstrates the challenging treatment of a patient with LCPD, who may present with components of both impingement and dysplasia.

DISCLOSURE STATEMENT: The authors have nothing to disclose.

Differences in Squatting Biomechanics in Individuals with Unilateral and Bilateral Adolescent Hip Dysplasia

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INTRODUCTION: Adolescent hip dysplasia (AHD) is an abnormal skeletal development that can lead to instability and/or subluxation of the hip joint [1]. Patients initially develop pain due to increased joint reactive forces which later leads to degenerative changes within the joint. Symptoms are often activity-related, but can progress to aggravation during sedentary activities [2], and in turn, may require surgical management. Understanding the clinical biomechanics of AHD may help with early detection and reduce complications later in life. Gait assessments may not be sensitive enough to display abnormalities in patients with AHD. Squatting, which requires adequate joint ROM, bilateral control of the lower limbs, and relies on the hips to effectively link the upper and lower body kinematics, is an activity which mimics activities of daily living which exacerbate symptoms, such as rising from a chair or getting into a vehicle [2]. The purpose of this study was to describe and compare how patients with unilateral (uAHD) and bilateral (bAHD) AHD complete a squat task.

CLINICAL SIGNIFICANCE: It has been shown that patients with uAHD often develop bilateral dysplasia [3]; understanding the differences that exist between these groups, may help with monitoring uAHD, allowing for earlier detection and treatment of the contralateral side.

METHODS: 24 patients with uAHD (age 17.5±3 years, height 161.4±9cm, and mass 62.8±16kg) and 34 patients with bAHD (age 16.7±4 years, height 164.5±10cm and mass 60.0±12kg) who were scheduled to undergo hip preservation surgery took part in this IRB approved study. Exclusion criteria included any secondary hip diagnosis or syndrome. Motion analysis data were collected and processed using a full body marker set (120Hz, VICON, Denver, USA) and two force plates (3000Hz, AMTI, Watertown, USA). Patients stood with the feet roughly shoulder width apart, on separate force plates. They were asked to raise their arms in front of them and squat to their perceived maximal depth, hold this position for 3 seconds and return to standing. The squat trial was divided into three phases: descent, hold and ascent [4]. Dependent variables included maximal trunk, hip, knee and ankle flexion as well as moments, impulse and power. Variables of the affected limb (surgical side for bAHD) were compared between the two groups. Ground reaction forces (GRF) in the z-direction were assessed as a ratio between the affected and unaffected limb (non-surgical side for bAHD) to assess asymmetry. Multivariate ANOVAs were run to compare differences between groups for each separate phase of the squat.

RESULTS: In all phases of the squat, maximum trunk flexion was greater in the uAHD compared to bAHD group (descent: p=0.002; hold: p=0.002; ascent: p=0.003), however, no significant differences were observed between hip, knee or ankle flexion. Maximal hip moments were increased in the uAHD group during descent (*p*=0.004) and ascent (*p*=0.04). A weak trend was observed between groups, showing greater hip power absorption (*p*=0.13) and work (*p*=0.09) in the uAHD group during descent. During the ascent phase, there was a significant increase in hip impulse (*p*=0.04) and a weak trend for increased hip work (*p* = 0.11) in the uAHD population. Values of significant and trending variables can be found in Table1. There were no significant differences in GRF ratios between groups.

Table 1: Values of trunk flexion, as well as, hip moments, power, work and impulse (mean±SD) in unilateral (U) and bilateral (B) patients with adolescent hip dysplasia, during descent, hold and ascent phases of a squat trial. * Significant differences ($p<0.05$) † trends ($p\leq0.1$)

		Trunk Flexion (°)	Hip Moment (Nm)	Hip Power (J/s)	Hip Work (J)	Hip Impulse (J)
Descent	U	29.58±13.26*	0.68±0.24*	-0.72±0.47†	-0.43±0.25†	0.67±0.41
	B	17.62±14.47*	0.55±0.22*	-0.55±0.37†	-0.32±0.24†	0.50±0.42
Hold	U	31.3±14.01*	0.47±0.21			
	B	19.56±13.74*	0.38±0.19			
Ascent	U	30.80±14.03*	0.70±0.2*	0.81±0.45	0.45±0.27†	0.49±0.23*
	B	19.15±13.90*	0.57±0.2*	0.76±0.39	0.35±0.23†	0.37±0.22*

DISCUSSION: Patients with uAHD exhibit different squatting biomechanics in comparison to patients affected bilaterally. While most maximal joint ranges may not differ statistically between groups, increased trunk flexion in uAHD results in greater hip flexion moments while descending into the squat and rising from the hold, placing greater stress onto the hip joint in comparison to bAHD. Similarities across joint ranges would suggest that any compensatory changes in squatting biomechanics are similar in both groups. A comparison of these ranges to healthy individuals without hip pathology would be beneficial to determine whether these values are typical. Interestingly, any significant differences in moments, work, power and impulse only occurred at the hip, showing this task does truly challenge the hip joint, and this task is an appropriate choice to assess hip biomechanics in this patient population.

REFERENCES:

1. Aronsson, D.D., et al. (1994) Pediatrics, 94(2):201-208
2. Sturridge, S., and Bankes, M. (2010) The Journal of Bone and Joint Surgery
3. Song, F.S., et al. (2008) Journal of Pediatric Orthopaedics, 28:173-176.
4. Stevens, W.R., et al. (2016) Journal of Biomechanics, *in review*.

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PILOT STUDY ON THE EFFECTS OF UPPER-LIMB LOSS AND PROSTHESIS USE ON LOCOMOTOR STABILITY

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INTRODUCTION

During steady-state walking, natural arm dynamics facilitate locomotor stability by minimizing body angular momentum through counterbalancing leg motion [1] and reducing trunk motion to constrain body center-of-mass (BCoM) excursion within the base of support [2]. Consequently, persons with upper-limb loss may experience reduced locomotor stability that may be dependent on prosthesis use. This pilot study investigated the effects of prosthesis use and matching inertial properties of the prosthetic limb to the sound limb on stability during walking.

CLINICAL SIGNIFICANCE

Use of a unilateral upper limb prosthesis may influence locomotor stability.

METHODS

Kinematics were measured using a 12 camera digital motion capture system (Motion Analysis Corp. (MAC), Santa Rosa, CA) and a modified Helen Hayes marker set [3] with additional trunk and arm markers. BCoM was estimated from an individual body segment model. A custom 'mock prosthesis' was designed to match the mass and inertial properties of the prosthetic limb to the sound limb (Figure 1). The mock prosthesis length, mass, and location of the center-of-mass were estimated using an algorithm based on established able-bodied anthropometric regression equations [4].

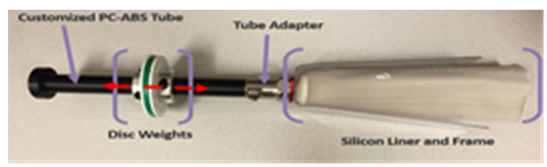

Figure 1. Mock prosthesis. Length is modified by replacing the plastic tube. Mass is modified through adding disc weights, secured in location by cuffs that allow translation of the center-of-mass.

Subjects walked over-ground on a level walkway at 3 self-selected speeds (slow, normal, and fast) under three prosthesis conditions: 1) without wearing a prosthesis, 2) wearing their customary prosthesis, and 3) wearing the mock prosthesis. At least five walking trials were collected and analyzed for each speed and prosthesis condition iteration.

Three-dimensional trunk rotations (relative to the global axes) were estimated using OrthoTrak software (MAC). Margin of stability (MoS) was estimated as the minimum distance between the lateral foot border (defining the base of support) and extrapolated BCoM (a velocity-weighted BCoM) positions [4]. Step width and variability of step width, length, and time were also calculated.

RESULTS

Two subjects with unilateral transradial (TR; 61yrs, 186cm, 90kg) and transhumeral (TH; 62yrs, 179cm, 111kg) limb loss participated in the study.

Trunk rotations at a single walking speed are displayed in Figure 2 (speed-matched at 1.3 m/s to data of 13 non-disabled control subjects (51±6yrs, 172±9cm, 74±15kg)). MoS was greater on the prosthetic and sound side for the TR and the TH subjects, respectively, but displayed little difference between prosthesis conditions. Minimal changes were seen in step width across conditions, but variability in step width, length, and time generally increased with use of a prosthesis (both customary and mock but with less clear individual trends).

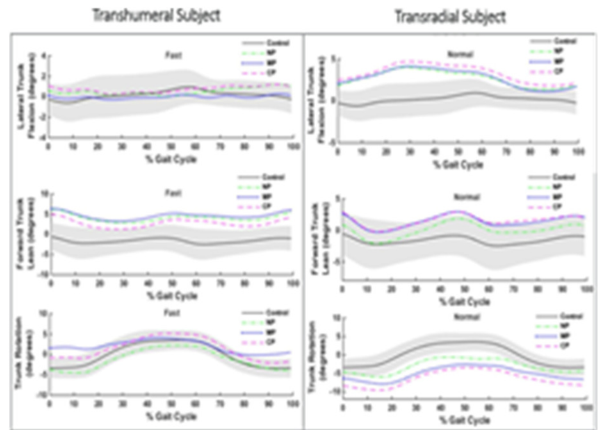

Figure 2. Frontal (top), sagittal (middle), and transverse trunk rotations (bottom) for both subjects. NP=no prosth, MP=mock prosth, CP=customary prosth.

DISCUSSION

These subjects with unilateral TR and TH limb loss displayed asymmetric trunk rotations with temporal profiles similar to controls. Trunk motion was minimally effected by prosthesis use and trends were not clear. Subjects displayed asymmetric MoS which aligned with the direction of asymmetry in lateral trunk lean, but was also not affected by the prosthesis. The side with greater MoS would suggest decreased opportunity for the BCoM to exceed the base of support and generate a risk to balance. Surprisingly, use of a prosthesis increased gait variability, suggesting reduced locomotor stability.

Individuals with unilateral upper limb loss walk with asymmetric trunk motion that is not impacted by prosthesis use. However, prosthesis use may result in reduced stability during walking and this should be further explored as it could affect the risk of falls. Data on eight additional subjects is currently being analyzed.

REFERENCES

1. Bruijn, S. et al. (2008) Gait and Posture, 27:455-462.
2. Shibukawa, M. et al. (2001) Engineering Med Biology Conference Proceedings.
3. Kadaba, M. et al. (1990) Journal Orthopedic Research, 8:383-392.
4. Winter, D. (2009) Biomechanics and Motor Control of Human Movement.

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DISCLOSURE STATEMENT

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Walking Sticks vs. Walker on Spine Kinematics in Adult Scoliosis Patients Pre and Post Surgery

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INTRODUCTION

Adult degenerative scoliosis (ADS) is associated with progressive and asymmetric degeneration of the disc and facet joints, which typically lead to stenosis.[1] By virtue of the narrowed spinal canal associated with the degeneration these patients frequently develop back pain, as well as leg pain, weakness, and numbness.[2] Patients with degenerative adult scoliosis demonstrate an altered gait pattern.[3] For many patients with ADS a walking aid is beneficial. Clinical experience has shown the use of walking sticks rather than a walker promotes a more upright posture. The walking sticks are beneficial pre-operatively not only in terms of deformity progression and line of sight, but also for patients postoperatively to help maintain surgical correction of their kyphotic deformities. The purpose of this study was to evaluate both the spatiotemporal and kinematic relationships of the lower extremities and spine during gait with walking sticks versus a walker in adult patients with degenerative scoliosis before and after their surgical intervention.

CLINICAL SIGNIFICANCE

Walking sticks may be a better assist with ambulation pre- and post- surgery in symptomatic ADS patients.

METHODS

Twelve subjects (Age: 66±6.8, H: 1.67±0.06 m, W: 81.5±25.9 kg, 6 females) with symptomatic degenerative scoliosis who have been deemed appropriate surgical candidates performed gait analysis under 3 testing conditions; 1. with walking sticks (WS), 2. with walker (WR), and 3. without any walking device (NW) on 2 occasions: 1. A week before the surgery (Pre), and 2. a month after the surgery (Post1). Fifty-one reflective markers (9.5 mm diameter) were incorporated to collect full body three-dimensional kinematics using 10 cameras (VICON, Denver, CO) at a sampling rate of 100 Hz. The patient walked barefoot at his/her self-selected speed along a 10 m walkway. Five trials were recorded during each session. Spine and lower extremity kinematics were measured and recorded. A 2 (time) X 2 (walking device) crossover mixed ANOVA design was used with Bonferroni Post Hoc analyses to determine differences in walking devices and times.

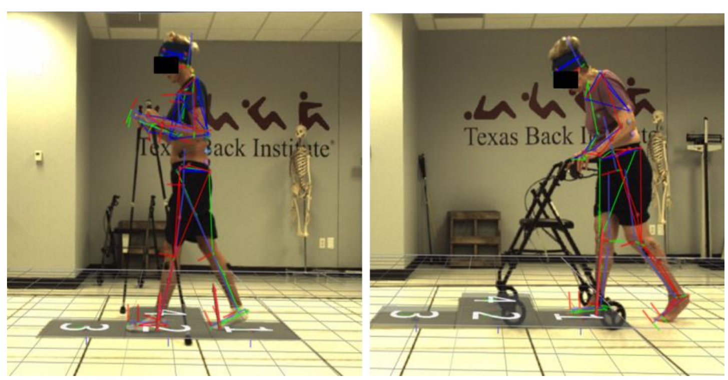

Figure 1. Gait with walking sticks and with walker in adult degenerative scoliosis patients

RESULTS

Spinal reconstruction did alter gait in ADS patients. This surgery improved the pelvic inclination (WS: 4.1° vs. WR: 0.5° vs. NW: 1.9°, $p<0.028$), lumbar (WS: 4.1° vs. WR: 4.1° vs. NW: -2.5°, $p<0.012$), neck (WS: 9.6° vs. WR: 8.4° vs. NW: 10.5°, p<0.001), head orientation (WS: 17.5° vs. WR: 26.6° vs. NW: 14.0°, p<0.012).

In the preoperative analysis, the use of walking sticks resulted in a significantly lessor pelvic inclination, (1.1°, p<0.028), lumbar (6.4°, p<0.012), neck (3.0°, p<0.021), head orientation (-6.3°, p<0.019) angles in comparison to the walker, but greater pelvic inclination, (1.8°, p<0.028), lumbar (1.7°, p<0.027), neck (2.0°, p<0.032), head orientation (1.2°, p<0.0143) angles in comparison to the walking without any device condition.

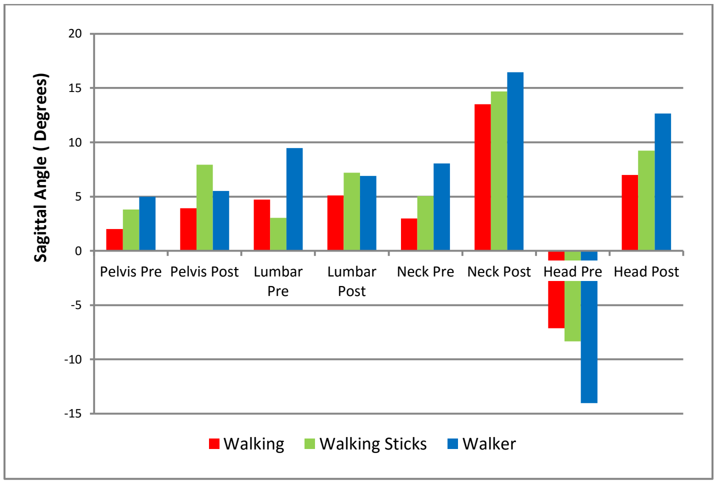

Figure 2. Spine kinematic angles with walking sticks, walker and without any walking device pre- and 1 month post-surgery in adult degenerative scoliosis patients

In the postoperative analysis, the use of walking sticks resulted in a significantly increased pelvic inclination, (2.4°, p<0.028), lumbar (0.2°, p<0.032), and smaller neck (1.9°, p<0.041), head orientation (3.4°, p<0.036) angles in comparison to the walker, but also increased pelvic inclination, (4.0°, p<0.029), lumbar (2.1°, p<0.033), neck (1.1°, p<0.041), head orientation (2.2°, p<0.032) angles in comparison to the walking without any device condition.

Those changes may not be clinical meaningful, but the overall change of the spine alignment is clinically noteworthy.

DISCUSSION

The clinical findings of less kyphotic posture with the use of walking sticks were verified through gait analysis in this study. There were clear improvements in sagittal kinematic parameters to support the findings of a more upright walking position. Whereas a walker forces patients into kyphosis, the higher grips of walking sticks allows for more upright posture and improved sagittal alignment. With preoperative walking stick training, surgical correction of deformity, and postoperative use of walking sticks, improvement in both sagittal parameters and kinematics as compared to a walker can be expected. We recommend the use of walking sticks to assist with ambulation in symptomatic adult degenerative scoliosis patients.

REFERENCES

1. Kotwal S, et al. *HSS J* **7**(3), 257-264, 2011.
2. Kotwicki. T, et al. *Adolesc Health Med Ther* **4**, 59–73, 2013.
3. Yang. JH, et al. *Eur Spine J* **22**(11), 2407–2413, 2013.

Case Study: Use of OpenSim Musculoskeletal Modeling to Optimize Orthotic Tuning in a Patient with Hemiplegic Cerebral Palsy.

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PATIENT HISTORY

An 8 year old child with a diagnosis of right hemiplegic CP, GMFCS level II presented for treatment with a chief complaint of right-sided toe walking, tripping on the right lower extremity and right knee hyperextension. The patient was born full-term. She sat at 12 months of age and began to walk independently at 18 months of age. Surgical history includes multiple instances of botulinum-A injections to the right hamstring and gastrocsoleus complex. Previous orthotic interventions were unsuccessful in improving her gait pattern.

CLINICAL DATA

Physical examination of the patient demonstrated range of motion limitations at the ankle, and decreased muscle strength and control of the right lower extremity; particularly the musculature distal to the knee joint. Right ankle dorsiflexion range of motion was found to be limited to -24 degrees (R1) and -20 degrees (R2) with the knee extended. Right hip extension, adduction and abduction muscles were graded 3+/5. Dorsiflexion, inversion and eversion of the right ankle musculature were graded 1/5 and plantarflexion strength was graded 2-/5. Observational gait assessment demonstrated right initial contact at the forefoot with knee hyperextension in terminal stance phase during barefoot walking.

MOTION DATA

Quantitative gait analysis was performed in barefoot and braced conditions two weeks following delivery of the right AFO using the Vicon motion capture system. Notable findings in barefoot walking include absent first rocker, persistent ankle plantarflexion and midstance knee hyperextension of the right lower extremity. On the left lower extremity, the collected kinematic data demonstrate increased hip and knee flexion in stance and swing phases of the gait cycle; as well as increased knee loading response during barefoot walking.

TREATMENT DECISIONS AND INDICATIONS

Data from the physical examination, observational and quantitative gait data assisted with the treatment planning for this patient. A solid ankle-foot orthosis (AFO) brace was recommended and fabricated in 20 degrees of ankle plantar flexion to accommodate for limitation in dorsiflexion range of motion. Heel posting and shoe wedging was added to the exterior of the brace to maintain a shank-to-vertical angel of 5 degrees of tibial incline. Following delivery of the brace and a period of accommodation, the patient presented to the Motion Analysis Laboratory for instrumented gait analysis using the Vicon motion capture system. Data was collected in barefoot walking and the right tuned AFO and footwear. OpenSim modeling using data obtained during three-dimensional movement analysis allowed for muscle-tendon length estimations in the two conditions (barefoot and braced).

OUTCOME

Comparison of barefoot and braced conditions revealed kinematic and kinetic gait changes including normalization of the right hip and knee range of motion during stance and swing phases of the gait cycle during braced walking. Kinetic changes included normalization of the knee extensor moment on the right side during braced walking (Figure 1). OpenSim modeling of the muscles of the lower extremity when using the customized and tuned AFO demonstrated normalized available muscle lengths of the right lower extremity; particularly the semimembranosus, gracilis, semitendinosus and medial gastrocnemius musculature. (Figure 2)

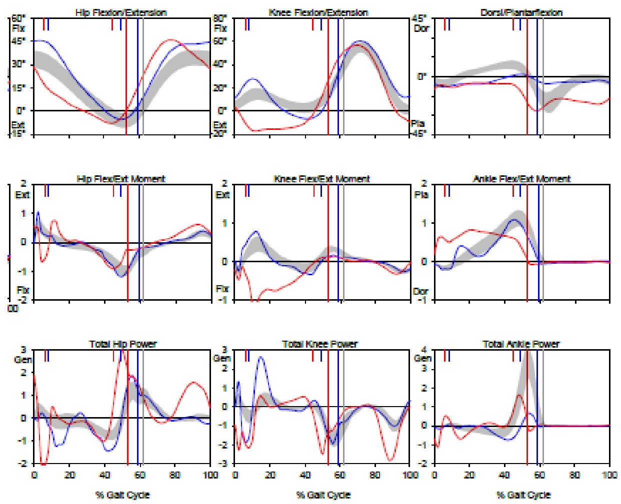

Figure1: Kinematic and kinetic data of right LE in barefoot (red) and braced conditions (blue)

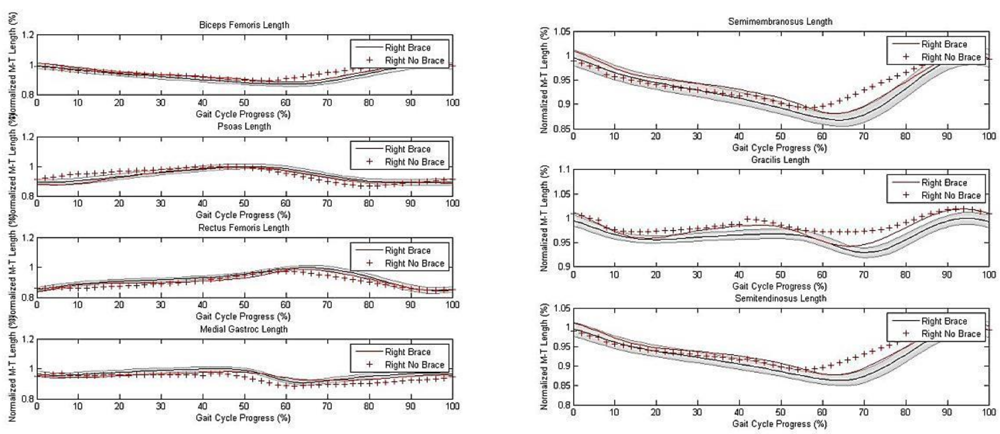

Figure 2: OpenSim muscle tendon modeling in barefoot (+) and braced (solid red) conditions

SUMMARY

For this child with right-sided hemiplegic CP and a right equinus gait pattern, use of tuned bracing resulted in improved knee and hip kinematic and kinetic patterns during walking. Furthermore, use of OpenSim modeling demonstrated improved available muscle lengths of the muscles proximal to the brace; particularly the hamstring musculature. For this patient with right hemiplegic CP, use of OpenSim muscle length modeling was useful in demonstrating improved available muscle lengths of the treated limb during walking in the customized orthotic prescription. Further assessment is needed to determine long-term results and generalizability to larger populations of children with CP.

DISCLOSURE STATEMENT

None of the authors have conflicts of interest to disclose.

Title: Protective stepping in people with MS: effects of a single bout of practice.

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INTRODUCTION: People with multiple sclerosis (MS) exhibit slow reactive steps, possibly contributing to falls [1, 2, 3]. However, whether reactive stepping can be improved in people with MS is unknown. Our aim was to determine whether people with MS can improve protective stepping through repeated exposure to external perturbations, and whether baseline cognitive performance may predict this improvement. We hypothesize that people with MS will improve stepping via practice and that participants with higher cognitive performance will exhibit the most pronounced improvements [4].

CLINICAL SIGNIFICANCE: Reactive postural control, including protective stepping, is a critical component of fall prevention after a loss of balance. Identifying 1) whether people with MS can improve protective steps, and 2) effective ways to improve these steps may reduce falls in this population.

METHODS: Protective stepping was elicited in 12 healthy adults and 4 people with moderate MS via repeated movements of the ground underfoot (i.e. postural perturbations) which required a reactive step. We measured changes in protective stepping over the course of 25 backward stepping perturbations. Margin of stability (MOS; distance between the stepping foot and center of

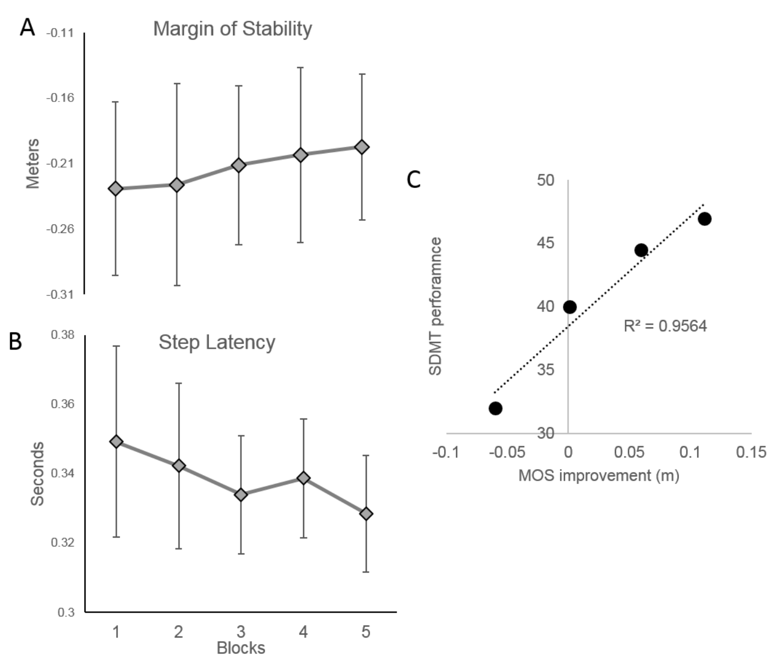

Figure 1: Changes in A) margin of stability (MOS), and B) step latency over the course of 25 backward stepping trials (5 blocks of 5 trials). Increased margin of stability and decreased step latency represent more efficient and faster steps, respectively. C) Relationship between improvement in margin of stability across training (x-axis) and performance on the symbol digit modality test (SDMT; y-axis). Although only 4 subjects have been collected, the positive relationship suggests that improved cognitive performance (measured by the SDMT) may predict improvements in performance over training.

mass at instance of first foot contact) and step latency (SLA; time to first foot off) were calculated via motion capture (Vicon Ltd; Centennial, CO) and imbedded force-plates. Cognitive function was also assessed in people with MS via the symbol digit modality test (SDMT). To date, we have collected data on 4 people with MS (mean[SD] of the European Database of Multiple Sclerosis (EDMUS) scale: 5[0.81]) and 12 healthy adults. Data collection on people with MS ongoing.

RESULTS: As reported previously, healthy adults improved backward stepping, as demonstrated by increased MOS and decreased step latency over the course of 1 day of protective stepping practice (MOS_start[m]: 0.11 - MOS_end: 0.15; p=0.001; SLA_start [ms]: 293 - SLA_end: 281; p=0.04). In people with MS, MOS and step latency were not significantly improved over practice (MOS_start[m]: -0.22; MOS_end: -0.19; p=0.51; SLA_start[ms]: 349 - SLA_end: 329; p=0.31), due in part to a currently small sample and high variability. However, as noted in figure 1 A & B below, subtle improvements were noted in both variables. Furthermore, although only 4 patients have been collected, cognitive capacity, measured by SDMT, is directly related to improvement in MOS over practice (figure 1c). In other words, people with better baseline SDMT scores exhibited more pronounced improvement in stepping performance over the course of practice. However, it is difficult to draw definitive conclusions based on this small sample, and additional data are currently being collected to better understand this trend.

CONCLUSIONS: Although further investigation is necessary, these preliminary findings suggest that improvements in protective stepping in people with MS are variable and likely less pronounced than healthy adults, and baseline cognitive capacity may be predictive of the degree of improvement in stepping.

REFERENCES:

1) Peterson DS, et al. (2016). Characterization of Compensatory Stepping in People With Multiple Sclerosis. Arch Phys Med Rehabil 2016;97:513-521. Huisinga
2) Huisinga JM, et al. (2014) Postural response latencies are related to balance control during standing and walking in patients with multiple sclerosis. Arch Phys Med Rehabil 2014;95:1390-1397.
3) Mansfield A, et al. (2013) Is impaired control of reactive stepping related to falls during inpatient stroke rehabilitation? Neurorehabil Neural Repair 2013;27:526-533.
4) Cai L, et al. (2014) Brain plasticity and motor practice in cognitive aging. Frontiers in aging neuroscience 2014;6:31.

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DISCLOSURE STATEMENT: The authors have no conflicts of interests to disclose.

IMPULSE AND MUSCLE ACTIVITY OF THE LEADING LIMB DURING BACKWARD AND LATERAL LUNGES AS WELL AS THE TRAILING LIMB DURING FORWARD LUNGE

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INTRODUCTION

Gluteus Maximus, Vasti, Gastrocnemius and Soleus are the extensor muscles suggested to provide support and forward progression in walking [1]. Physical therapists prescribe lunge exercises to activate their clients' extensor muscles. Although kinetics and muscle recruitment of the leading/initiating limb of the forward lunge (FL) have been studied [2-8], the trailing limb of the forward lunge has not. Some have compared the initiating limb during the FL and Lateral Lunge (LL) [2,7,8] while others compared the FL to the Backward Lunge (BL)[5], but we found no publications comparing all three lunges. Furthermore, those studies of muscle activation used surface electrodes which are prone to cross talk from adjacent muscles. The purpose of this study was to compare the impulse and muscle activity of the initiating limb during BL and LL as well as the trailing limb during the Forward Lunge (FL_TL).

CLINICAL SIGNIFICANCE

Data from this study will assist the physical therapist in choosing the most appropriate lunge exercise to create a demand for specific extensor muscles targeted for the client.

METHODS

Twenty-nine young adults with no history of recent lower limb injury were studied with intramuscular electromyography (WEMG) of the Soleus (S), Medial Gastrocnemius (MG), Vastus Lateralis (VL) and Lower Gluteus Maximus (LGMax) muscles and for hip, knee and ankle joint impulse of the initiating limb during the LL and BL and of the FL_TL. WEMG data were acquired using a band-pass filter of 20-1000 Hz, sampled at 2000 Hz. RMS values were normalized as percent maximum voluntary isometric contraction (%MVIC). Joint impulse was calculated by link segment modeling using an 8-camera optoelectric system and forceplates. Three trials were performed per exercise. Data were analyzed by Friedman's and Wilcoxin signed-rank tests with medians reported. The study obtained IRB approval from California State University, Long Beach.

RESULTS

All lunges produced significantly higher extension impulses at the knee than at the ankle but knee extension impulse did not differ across lunges, averaging 1.94 Nm/Kg*s. Likewise, all lunges produced significantly more VL muscle activity than MG. Those exercises studying the initiating limb (BL and LL) also recruited significantly more VL than S activity. In the study of the trailing limb (FL_TL), where ankle extension impulse was greatest (1.68 Nm/kg*s), S was equal to both VL and MG activity.

LL was most effective in recruiting extensor muscles (S=20.2, VL= 47.2, LGMax=18.5 % MVIC). MG activity did not differ across lunges, averaging 10.0 %MVIC. LL was also the only lunge with a hip extension impulse (2.01 Nm/Kg*s) whereas BL and FL_TL produced flexion impulses.

DISCUSSION

All lunge exercises targeted knee extension kinetics and VL muscle activity. The LL produced extension impulses at all 3 joints and recruited the most activity of the VL, S and LGMax. The FL_TL had highest ankle extension impulse.

Farrokhi et al, [7] found that trunk forward lean during FL increased hip extension moment when compared to an erect trunk. Our results support this finding as our only exercise with a trunk forward lean, LL, was the only lunge that produced a hip extension impulse and recruited LGMax EMG.

Physical therapists may use these lunge exercises to recruit the client's knee extensors but should consider the LL over BL or FL_TL for LGMax recruitment and for higher recruitment of VL and S. If the client has knee disability prohibiting high VL activation, the TL_FL may be used to target calf muscles.

REFERENCES

1. Pandy, MG. et al. (2010) Annual Review of Biomedical Engineering, 12:401-33.
2. Flanagan, SP. et al. (2004) Journal of Strength Condition Research, 18(3):599-605.
3. Farrokhi, S. et al. (2008) Journal of Orthopedic and Sports Physical Therapy, 38(7):403-409.
4. Riemann, BL. et al. (2012) Journal of Athletic Training, 47(4):372-378.
5. Comfort, P. et al. (2015) Journal of Athletic Training, 50(10):1011-1018.
6. Jonhagen, S. et al. (2009) Scandinavian Journal of Medical Science and Sports,19(4):561-568.
7. Begalle, RL. et al (2012) Journal of Athletic Training, 47(4):396-405.
8. DiStefano, LJ. et al (2009) Journal of Orthopedic and Sports Physical Therapy,39(7):532-540.

DISCLOSURE STATEMENT

We have no conflicts of interest to disclose.

Biofeedback Results in Gait Characteristics Similar to Controls with Decline Walking after Total Knee Arthroplasty

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INTRODUCTION

Total knee arthroplasty (TKA) is the standard surgical procedure for managing chronic pain and disability related to advanced knee osteoarthritis [1]. TKA is effective at reducing pain and improving patient-reported outcomes, however persistent gait deficits exist years after surgery [1, 2]. Abnormal gait mechanics have been shown to persist after total knee arthroplasty (TKA) and do not return to the same levels as healthy controls [3, 4]. Aberrations consistently increase during physically demanding tasks like stairs or decline walking [3]. Knee kinetic biofeedback (FB_{RT}) might improve chronic compensation patterns following TKA. Correcting faulty joint mechanics could impact muscle disuse in the surgical limb and compensatory overloading of the non-surgical limb [3, 4]. The purpose of this project was (1) to determine if gait characteristic differences existed between healthy matched-controls and TKA patients and (2) to determine if TKA patients' gait characteristics could be normalized, relative to controls, with use of knee kinetic FB_{RT} during a more physically demanding task such as decline walking.

CLINICAL SIGNIFICANCE

FB_{RT} has been used in rehabilitation to address chronic compensatory strategies following TKA [1]. However, nothing to date has looked at using FB_{RT} during decline walking. Inter-limb sagittal knee moment was implemented for FB_{RT} knowing chronic reductions in knee moments are commonly found following TKA [1, 2]. FB_{RT} provides a novel intervention that has potential to improve upon an improper learned behavior developed prior to TKA to avoid pain, but continues to exist following successful pain reduction after surgery. This novel mode of retraining could assist in correcting faulty movement patterns and improve gait characteristics to similar levels as matched-controls. Although FB_{RT} has been shown to be effective at improving knee kinetics during level walking [1], it is not clear if these compensatory strategies can be mitigated through FB_{RT} (knee moments) during decline walking. It is vital to evaluate more demanding mobility tasks, relative to level walking, in person following TKA knowing these are necessary tasks in returning persons back to prior level of function. Determining how these deficits influence compensatory movement strategies during decline walking will provide the necessary framework to develop rehabilitation interventions that can be implemented into a longitudinal intervention trial.

METHODS

Prospective study design including only uncomplicated primary TKA patients and matched-controls. Groups were similar in age ($p = 0.89$), sex ($p=0.66$) and body mass index ($p=0.98$). Clinical outcomes, using PROMIS physical function computerized adaptive testing (PF-CAT) and numeric pain rating scale (NPRS) scores were compared between groups. All gait characteristics were collected on a 10° decline duel-belt instrumented treadmill (Bertec, Columbus, OH) at a constrained velocity of 0.8 m s^{-1}. Between group differences in vertical ground reaction force (vGRF), peak knee joint angle (PKJA), total support moment (TSM) and knee joint moment (KJM) were compared using t-tests. FB_{RT} using knee joint moments were calculated using Visual3D (C-Motion, Inc., Germantown, MD) and displayed to the patient during the treadmill trials (Fig. 1). The kinetic and kinematics variables were collected in Nexus 2.1 (Vicon, Oxford, UK). Using inverse dynamics through rigid body analysis (Visual3D) based on the anthropometrics of the participant and the ground reaction forces from the force plates and joint angles of the lower limbs, the individual joint moment contributions were derived within all trials.

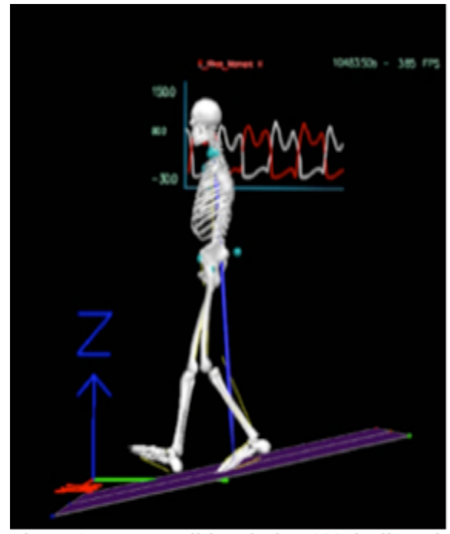

Figure 1. FB_{RT} condition during 10° decline trial.

DEMONSTRATION

Relative to healthy controls, TKA patients demonstrated significant differences in PKJA ($p=0.05$), KJM ($p<0.01$) and TSM ($p<0.001$) during the decline walking task. However, TKA patients with FB_{RT} were able to normalize gait characteristics similar to healthy controls in PKJA ($p=0.57$) and KJM ($p=0.20$). Mean PROMIS PF-CAT T-score was 52.3 in the control group and 46.3 in the TKA group ($p<0.01$). Mean NPRS score was 0.2 in the control group and 0.8 in the TKA group ($p=0.14$).

SUMMARY

TKA patients with FB_{RT} were able to make immediate changes in gait characteristics to levels similar to healthy matched-controls during a complex and physically demanding task of decline walking. Despite good patient-reported outcomes and minimal knee pain, patients without FB_{RT} displayed significantly different gait characteristics compared to healthy matched-controls.

REFERENCES

1. Zeni J et al. (2003) JOSPT, 43(10):715-726.
2. Mizner R et al. (2005) Journal of Orthopaedic Research,23(5):1083-1090.
3. Komnik I et al. (2015) Gait & Posture, 41(2):370-377.
4. Levinger P et al. (2013), The Journal of arthroplasty, 28(6):994-999.

DISCLOSURE STATEMENT

Not applicable

THE CHANGE OF THORACIC SPINE ANGLE AND LUMBAR SPINE ANGLE DURING SPEED-UP RUNNING

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INTRODUCTION

Some runners have low back pain (LBP) such as lumbar spondylolysis. The seven percent of runners have lumbar spondylolysis [1]. The causative factor of lumbar spondylolysis is repeating lumbar extension with rotation [2].

During running, they extended their spine 14° and tilted their pelvic 20° anteriorly [3]. However, there were no reports for the change of women's thoracic spine and lumbar spine angle during speed-up running. The purpose of this study was to clarify the change of the thoracic and lumbar spine motion during speed-up running.

CLINICAL SIGNIFICANCE

We thought that clarifying the change of thoracic and lumbar spine angle could be useful for prevention of LBP.

METHODS

We recruited seven college track team women runners (age, 19.4 ± 1.4years; height, 159.0 ± 7.9 cm; weight, 47.8 ± 4.1 kg; BMI 18.8 ± 0.5 kg/m^2). We measured running motion using a three-dimensional motion analysis system with 250 Hz (Qualisys track manager, Qualisys AB., Sweden). We used a treadmill (IGNIO R-16S, Alpen Co.,Ltd. JAPAN) and change the speed from six to nine and 12 km/h. We placed 68 spherical markers on each anatomical landmark and calculated the angle of the thoracic and lumbar spine.

We collected data for the following four events: right foot contact, right toe off, left foot contact, and left toe off. We converted the 10 cycle's data into 100%. We used a 1-way ANOVA with post hoc test of bonferroni to compare the thoracic and lumbar spine angle during speed-up running. The level of significance was set at $p < 0.05$.

RESULTS

Except for the left foot contact time, there were significant differences between time parameters in 6 km/h and 9 or 12 km/h ($p < 0.05$, Table 1). Except for the sagittal range of thoracic spine angle, all the range of thoracic and lumbar spine angle significantly increased during speed-up running ($p < 0.05$, Fig. 1). As speed-up running, their thoracic spine was extended, bended ipsilaterally, and rotated contralaterally in the early stance phase. In this phase, their lumbar spine of extension angle was decreased and contralateral bending angle was increased. In the late stance phase, their lumbar spine of extension and contralateral rotation was increased.

Table 1: Change of each time parameter during speed-up running

	Stance phase		Swing phase		R toe off		L foot contact		L toe off	
	mean	SD	mean	SD	mean	SD	mean	SD	mean	SD
6 k/h	84.8	7.9	15.2	7.9	43.2	5.7	50.8	1.5	92.5	3.7
9 k/h	69.8	8.5	30.2	8.5	35.4	6.0	51.3	3.0	85.7	5.1
12 k/h	66.1	8.0	33.9	8.0	33.5	5.1	49.5	6.4	82.0	4.0

DISCUSSION

Participants moved their lumbar spine more than their thoracic spine during speed-up running. They extended their hip with anterior inclination of pelvis [3], which would increase lumbar extension. To maintain the lateral balance during speed-up running, participants would increase the spine angle of lateral bending. To increase the running velocity, participants would increase the spine angle of rotation.

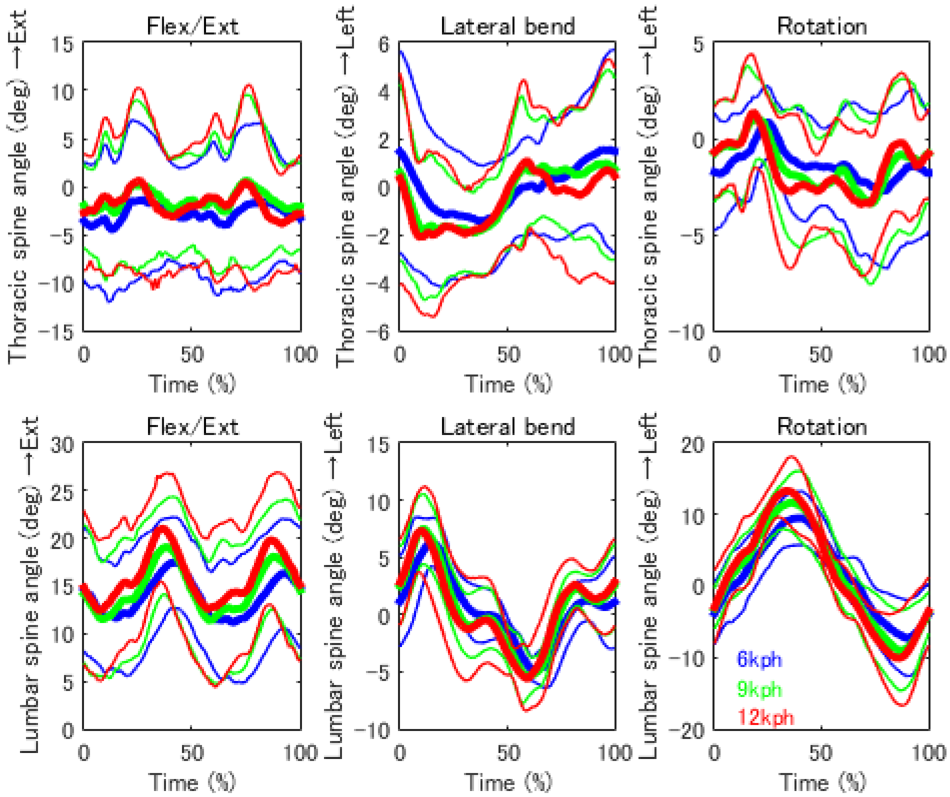

Figure 1: Mean (thick lines) and standard deviation (thin lines) for thoracic and lumbar spine angle during speed-up running. Blue lines, 6 k/h; green lines, 9 k/h, and red lines, 12 k/h.

The mechanical stress causes LBP during trunk extension with rotation [2]. It would be important paying attention in the late stance phase to prevent LBP because runner increased the extension and rotation angle of lumbar spine during speed-up running.

REFERENCES

[1] Soler T, Calderon C. The prevalence of spondylolysis in the Spanish elite athlete. Am J Sports Med 2000; 28(1):57-62.

[2] Sairyo K, Katoh S, Sasa T, Yasui N, Goel VK, Vadapalli S, Masuda A, Biyani A, Ebraheim N. Athletes with unilateral spondylolysis are at risk of stress fracture at the contralateral pedicle and pars interarticularis: a clinical and biomechanical study. Am J Sports Med 2005; 33(4):583-90.

[3] Rendos NK, Harrison BC, Dicharry JM, Sauer LD, Hart JM. Sagittal plane kinematics during the transition run in triathletes. J Sci Med Sport 2013; 16(3):259-65.

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DISCLOSURE STATEMENT: All authors have no conflicts of interest to disclose.

COMPARISON OF KNEE JOINT FORCE AND MOMENT DURING SHORT AND MIDDLE TURNS OF CARVED SKI USING WEARABLE MOTION ANALYSIS SYSTEM

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INTRODUCTION

Skiing is one of the most popular winter sports in the world. According to the US consumer product safety commission, the number of people treated for winter sports-related injuries in 2014 is more than 290,000 and the biggest amount of injuries (114,000 injuries) are from snow skiing. Even though the equipment, such as helmets and bindings, help to improve safety during skiing, injury risks of the lower extremity are still high [1]. Especially, knee injuries are the most common problem in professional skiers [2]. In order to prevent the injuries, it is necessary to investigate injury risk in various motions during carved ski. In this study, we captured skiing motion during short-turn and middle-turn and investigated joint force and moment in knee using multi-scale computer simulation of human musculoskelectal system.

CLINICAL SIGNIFICANCE

Three-dimensional knee joint force and moment were evaluated based on the data obtained from wearable motion analysis system. Excessive joint force or moment might result in the osteoarthritis and ligament injuries, investigation of joint force and moment could provide fundamental information to understand injury risks during sports activities such as skiing.

METHODS

Seven male certified ski coaches (age: 39.5±14.5 yrs, height: 179±4 cm, weight: 86.5±8 kg, career: 18.5±12.5 yrs) were participated in this study. All subjects used their own equipment after taking the wearable motion capture system, which consists of inertial measurement unit sensors (MVN, Xsens Technologies, Netherlands) and plantar pressure sensors (Pedar X, Novel, Germany). All motions and plantar pressure were captured during skiing with both short- and middle-turns.

Joint kinematics and kinetics of the right lower extremity were analyzed during left turns using short- and middle-turns because skiers use the opposite legs of the rotational directions. Here, the cross point between center of mass (COM) and ski's mean position was determined as start and end points of the turn [3]. The turns were normalized to 100% from the determined start point to the end point. The mean values of joint angles, forces, and moments in the hip, knee and ankle were then calculated by the validated inverse dynamics technique.

RESULTS

The pattern of hip and knee flexion-extension angles showed different between short- and middle-turns. During middle-turn, the mean joint angle of hip and knee were about 10° and 6° less flexion angle than those during short-turn, respectively. However, the patterns in abduction- adduction and internal-external rotation were quite similar in two turns (Figure 1). In addition, the ranges of motion in all rotation directions were also similar in two turns.

Higher joint forces and moments were predicted in all joints during middle-turn. Especially maximum anterior-posterior force was 2.6 times higher during middle-turn than during short-turn. Also, 2.4 times higher internal moments at the knee joint were predicted during middle-turn compared to the moment during short-turn. During short-turn, only extension moment was predicted, however, during middle-turn, flexion moment was also predicted that about 34 Nm at the end of the middle-turn.

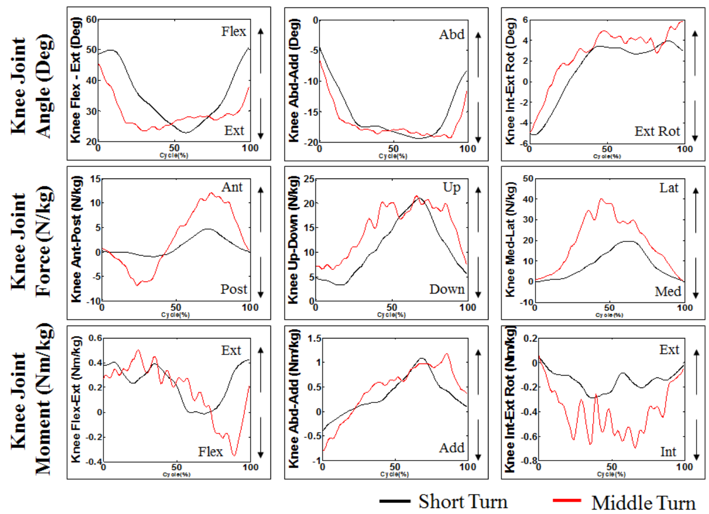

Figure 1: Changes in joint angle, forces, and moments at the knee joint during short- (black) and middle-turns (red)

DISCUSSION

Even though slight difference of joint kinematics between short-turn and middle-turn were predicted, higher joint forces and moments during middle-turn were predicted compared to the forces and moments during short-turn. Because these higher joint forces and moments could result in the osteoarthritis and ligament injuries at the knee joint, injury risk of the knee joint during the middle-turn may be higher than during the short-turn on the aspect of the osteoarthritis and ligament injuries. It might be suggested using the short-turn for the ski player in the rehabilitation process from knee joint injury as a safer turn.

REFERENCES

1. Klous, M. et al. (2012) Journal of Applied Biomechanics, 28(6):655-664.
2. Bere, T. et al. (2011) The American Journal of Sports Medicine, 39(7):1421-1429.
3. Supej, M. et al. (2003) Kinesiologia Slovenica, 9(1):11-17.

ACKNOWLEDGMENTS

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DISCLOSURE STATEMENT

All authors have no conflicts of interest to disclose.

COMPARISON OF LINEAR AND NON-LINEAR APPROACHES TO ESTIMATING SCAPULAR MOTION

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INTRODUCTION

Accurate assessment of scapulothoracic (ST) kinematics is important for understanding shoulder function [1], but traditional 3D motion capture has failed to provide accurate estimates of scapular kinematics throughout a full range of arm motion. For example, the most commonly utilized approach to estimating ST motion is limited to humerothoracic (HT) elevation angles of 120 degrees or less [2]. An approach that aims to remove this limitation utilizes the relationship between humeral orientation, acromion process (AP) displacement, and ST orientation to develop individualized equations to estimate ST orientation throughout a full range of HT motion. However, it is unknown whether the scapula moves in a linear or non-linear manner with respect to HT orientation and AP displacement. The goal of this study was to compare both linear and non-linear approaches to estimating ST kinematics among typically developed individuals. Individualized algorithms that estimate dynamic scapular orientation based on measured humeral orientations and AP displacements were developed using both linear and non-linear mathematical approaches. The results of the mathematical approaches were compared to a validation standard collected with biplane fluoroscopy and 2D to 3D image registration [3].

CLINICAL SIGNIFICANCE

This study will identify a mathematical approach for the clinical assessment of dynamic scapular orientation. This new technique is expected to drive future work in musculoskeletal modeling, clinical trials, and basic research that will add further insight into healthy and impaired shoulder function.

METHODS

Individualized multiple linear regression (LR) algorithms and artificial neural networks (NNs) were developed for nine healthy adult shoulders. Algorithms and networks were used to predict ST kinematics, and predicted kinematics were compared to kinematics obtained from fluoroscopy to determine the accuracy of each mathematical approach.

LR algorithms and NNs to estimate scapular orientation from AP displacement and humeral orientation were created for each subject using motion capture data (Motion Analysis Corp, Santa Rosa, CA) from eleven static positions: neutral, abduction, hand to mouth, hand to neck, elevation, forward reach, extension, hand to spine, external rotation, internal rotation, and flexion. Humeral orientation was calculated as HT helical angles, and AP displacement was calculated as the displacement of the AP marker along the Y axis (inferior/superior) and the X axis (anterior/posterior) of the trunk. The predicted values were scapula orientation, calculated as resolved ST helical angles. Three LR equations and three NNs were developed for each individual to estimate the X (upward rotation), Y (external rotation), and Z (anterior tilt) components of ST orientation. The LR algorithms and the NNs were applied to dynamic trials in which the individual moved continuously through the eleven positions listed above. The accuracy of the mathematical approaches was determined by correlations and average differences between the predicted ST angles and the ST angles measured using fluoroscopy.

RESULTS

Both mathematical approaches produced favorable results (Table 1). However, the average differences for each axis were lower and the average correlations for each axis were higher with the LR algorithms. Figure 1 shows the ST motion of a representative subject as estimated with LR algorithms and NNs.

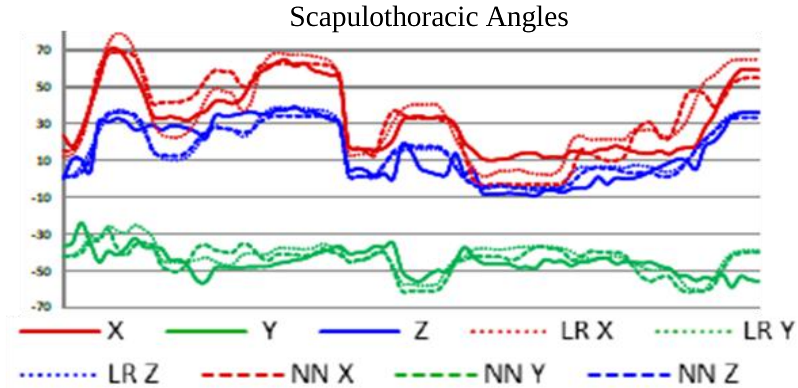

Figure 1: ST angles (degrees) during dynamic trials compared to the validation standard for one representative

Table 1: Linear regression (LR) and Neural Network (NN) result averages for all subjects (mean (SD)).

	ST X Diff	ST Y Diff	ST Z Diff	ST X Corr	ST Y Corr	ST Z Corr
LR	7.2° (1.4°)	7.4° (1.4°)	6.5° (1.8°)	0.92 (.03)	0.46 (0.28)	0.87 (0.01)
NN	8.8° (1.2°)	8.0° (1.8°)	7.7° (2.6°)	0.85(.05)	0.32 (0.31)	0.78 (0.16)

DISCUSSION

Although both mathematical approaches were able to estimate ST angles throughout the entire motion, the LR technique estimated ST angles that resulted in higher correlations with and smaller average deviations from the fluoroscopy measures. These results confirm that the relationship between HT motion, AP displacement and ST orientation is linear. The LR technique is also less complex, making it easier to implement in a clinical environment. It takes less than thirty minutes to collect the necessary motion capture data, and less than thirty seconds of computing time to generate the three LR equations for one subject, making LR algorithm development a fast, non-invasive, and clinically accurate tool for determining dynamic scapular orientation.

REFERENCES

1. Bourne, D. et al. (2004) Ann Biomed Eng, 39:777-785.
2. Karduna, A. et al. (2001) Journal of Biomech Eng, 123(2):184-190.
3. Nicholson, K. et al. *GCMAS Annual Conference Proceedings*, 2014.

DISCLOSURE STATEMENT

The authors have no conflicts of interest to declare.

Keynote #2

GCMAS PAST/PRESENT/FUTURE

THURSDAY, 25 May 2017
11:00A – 12:00PM

Michael D. Aiona, MD
Chief of Staff
Director Motion Analysis Center
Director Clinical Research Department
Shriners Hospitals for Children - Portland
Portland, OR

Michael D. Aiona, MD completed his orthopaedic surgery residency at the University of Utah Medical Center followed by a year of fellowship training in pediatric orthopaedics at the Shriners Hospitals for Children in Greenville, SC. Dr. Aiona worked at the Shriners Hospitals for Children in Lexington, KY for two years. Since 1986, he has worked at Shriners Hospitals for Children-Portland and now serves as the Chief of Staff. He is the Director of the Movement Analysis Laboratory and Clinical Research Department. His clinical expertise is in managing children with Cerebral Palsy and myelodyspasia, limb length inequality, club feet, lower extremity deformities and gait abnormalities.

Dr. Aiona serves as Associate Editor of Gait and Posture and is a reviewer for JBJS, JPO, and DMCN and has or is serving on several committees within the AACPDM.

Podium Session #5

PEDIATRIC GAIT/CEREBRAL PALSY II

MODERATED BY: **Sherry Backus, PT:** Clinical Supervisor, Leon Root MD,
Motion Analysis Laboratory**,**
Hospital for Special Surgery**,** New York, NY

Frank Chang, MD: Medical Director, Center for Gait and Movement Analysis
Children's Hospital Colorado**,** Denver, CO

1. **The Biomechanics of Running in Children**
   Devin Kelly, Corinna Franklin, Kevin Cooney

2. **Knee Joint Loading During Gait in Independently Ambulatory Children with Spina Bifida**
   Tishya Wren, Nicole Mueske, Deirdre Ryan, Katherine Steele

3. **Utilization of The Gait Profile Score (GPS) And The Gait Variable Scores (GVS) to Quantify Changes in Gait with Disease Progression in Boys with Duchenne Muscular Dystrophy (DMD)**
   Susan Sienko, Cathleen Buckon, Anita Bagley, Eileen Fowler, Kent Heberer, Loretta Staudt, Craig McDonald, Michael Sussman

4. **Ankle Dorsiflexion Differences Between Lunge Test and Gait in Patients with Charcot-Marie-Tooth**
   Kelly Pogemiller, Kristan Pierz, Matthew Solomito, Sylvia Õunpuu

5. **Rectus Sparing Approach to Periacetabular Osteotomy in Adolescents Preserves Hip Flexion Strength at Short Term Follow-up**
   Kirsten Tulchin-Francis, David A. Podeszwa, Adriana De La Rocha, Daniel J. Sucato

6. **Does Hip Abductor Function Increase After a Femoral Derotation Osteotomy in Individuals with Cerebral Palsy?**
   Elizabeth Boyer, Tom Novacheck, Michael Schwartz

7. **Excessive Femoral Anteversion Does Not Cause Excessive Anterior Pelvic Tilt**
   Michael Schwartz, Tom Novacheck, Elizabeth Boyer

8. **The Long-term Outcome of Pelvic Asymmetry During Gait in Children with Cerebral Palsy Following Unilateral Femoral Derotation Osteotomy**
   Lucio Perotti, Chris Church, Robert Dina, Nancy Lennon, John Henley, Julieanne Sees, Freeman Miller

THE BIOMECHANICS OF RUNNING IN CHILDREN

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INTRODUCTION

The biomechanics of running have been well studied in adults [1]. However, there are limited data available for use in clinical interpretation of atypical running patterns in children. Hypothesis-driven studies in the literature have explored biomechanics of running in children with resultant data provided to answer a specific research question including mechanical work at different speeds and chosen step frequencies, shock attenuation, and differences between overground and treadmill running [2-4]. Few have made distinctions between children of different ages and none have provided a complete, comprehensive dataset for clinical use. Therefore the purpose of this study was to determine the kinematics and kinetics of running in healthy children to serve as a pilot database for clinical comparison and to determine differences in these variables related to children's age.

CLINICAL SIGNIFICANCE

This study will enhance routine clinical analysis of atypical running patterns in children due to orthopedic or neuromuscular pathology by providing comparative kinematic and kinetic data in healthy children. It will also provide data on how running kinematics and kinetics may change as children age.

METHODS

Twenty two healthy, typically developing participants were separated into two groups by age. Group 1 (n=11; 4 M, 7 F) ages 4 to 9 (Mean:6.82 ;SD:1.78) and Group 2 (n=11; 6 M, 5F) ages 10 to 18 (Mean:13.64 ;SD:2.87). Participants visited the Motion Analysis Center at Shriners Hospital for Children, Erie on one occasion. The experimental procedures and protocol were explained. Informed consent including parental consent and youth assent were obtained as approved by the local IRB. General anthropometrics were taken. Three dimensional kinematic and kinetic data were collected at 120 and 1200 Hz respectively. Next, the participants were asked to complete 5 successful running trials per foot. A successful trial was defined by the foot striking the center of one force platform as the participant ran the length of the laboratory (13.72 m) barefoot at a self-selected velocity. Group differences were determined according to Cohen's d effect sizes with a value of d greater than 0.80 deemed clinically significant [5].

RESULTS

Spatiotemporal parameters were compared between groups (Table 1). Kinematic and kinetic data between groups were also compared. Differences in a subset of average peak values are presented in Table 2.

Table 1: Spatiotemporal parameters. Mean (Standard Deviation). * = clinically significant difference according to Cohen's d effect size.

	Group 1	Group 2	Effect Size
Preferred Velocity (m/s)	3.63 (0.34)	3.49 (0.50)	.33
Cadence (steps/min)	228.02 (25.96)	184.91 (16.08)	2.0*
Step Time (s)	0.27 (0.03)	0.33 (0.03)	2.0*
Step Length (m)	0.97 (0.14)	1.14 (0.19)	1.02*
Step Width (m)	0.05 (0.01)	0.05 (0.02)	0

Table 2: Average peak angles and internal moments. Mean (Standard Deviation). * = clinically significant difference according to Cohen's d effect size.

	Group 1	Group 2	Effect Size
Hip Extension (degrees ▫)	5.27 (6.41)	8.75 (5.02)	0.60
Knee Flexion (▫)	103.29 (14.94)	93.54 (12.71)	0.70
Ankle Dorsiflexion (▫)	11.29 (3.79)	17.02 (3.59)	1.55*
Ankle Plantarflexion (▫)	35.98 (4.82)	30.83 (7.41)	0.82*
Hip Extension Moment(Nm/kg)	1.98 (0.56)	1.63 (0.46)	0.68
Knee Extension Moment(Nm/kg)	1.05 (0.25)	1.69 (0.43)	1.82*
Ankle Plantarflexion Moment (Nm/kg)	2.17 (0.37)	2.50 (0.45)	0.80*

DISCUSSION

Results of this study show a number of differences in the running gait of younger (Group 1) versus older (Group 2) children. Although self-selected velocity was similar between groups, how each group achieved this velocity differed. The younger group had increased step rate and decreased step length as compared to the older group. This is a similar finding to spatiotemporal parameters observed during children walking [6]. Peak ankle dorsiflexion and plantarflexion differed between groups. The younger group exhibited a higher degree of plantarflexion while the older group exhibited a greater degree of dorsiflexion. There were a number of kinetic differences between groups including knee extension moment, and ankle plantarflexion moment. This suggests that as children develop some moments increase. Differences observed between younger and older children in this study show the importance of age specific kinematic and kinetic normative data for clinical comparison and decision making. Further, findings support future pursuit of a larger database for the biomechanics of running in children.

REFERENCES

1. Novackeck, T.F. (1998) Gait and Posture, 7:77-95.
2. Schepens et al. (1998) Journal of Physiology, 509(3):927-940.
3. Mercer et al. (2010) Journal of Athletic Training, 45(3):259-264.
4. Rozumalski et al. (2015) Human Movement Science, 39:200-211.
5. Cohen, J. (1988) Statistical power analysis for behavioral sciences, pp. 21-44.
6. Sutherland et al. (1980) The Journal of Bone and Joint Surgery, 62(3):336-353.

KNEE JOINT LOADING DURING GAIT IN INDEPENDENTLY AMBULATORY CHILDREN WITH SPINA BIFIDA

Tishya A. L. Wren[1], Nicole M. Mueske[1], Deirdre D. Ryan[1], and Katherine M. Steele[2]

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INTRODUCTION: Children with spina bifida have altered walking patterns due to varying degrees of weakness in trunk and lower extremity muscles. These walking patterns can lead to pain and possibly premature joint degeneration. The purpose of this study was to use gait analysis and musculoskeletal modeling to examine knee joint loads applied to the tibia in children with myelomeningocele, the most severe type of spina bifida. We hypothesized that knee joint loads would be elevated in children with myelomeningocele due to abnormal gait patterns.

CLINICAL SIGNIFICANCE: Our results suggest that knee joint loads tend to be reduced, rather than elevated, in independently ambulatory children with spina bifida, likely due to reduced walking speed. This might lessen the risk of developing joint problems and osteoarthritis due to overloading, but could also contribute to lower bone mass and increased fracture risk.

METHODS: 31 independently ambulatory children with myelomeningocele (14 female; mean age 10.0 years, SD 2.2, range 6-13; 15 sacral, 5 low lumbar, 11 mid lumbar) and 16 controls (7 female; mean age 10.1, SD 1.7, range 7-12) underwent motion analysis testing walking at a self-selected speed with or without braces and shoes as needed. Retro-reflective markers were placed on the trunk and lower extremities following the Plug-in-Gait model with patella markers instead of thigh wands. Data were collected at 120 Hz by an 8-camera motion capture system (Vicon, Oxford, UK). Synchronized ground reaction force data were collected by four floor-embedded force plates (AMTI, Watertown, MA) at 2520 Hz. Knee joint reaction loads were calculated from the motion analysis data using OpenSim [1]. First, a musculoskeletal model of the full body less arms [2] was scaled to each participant. Joint angles were calculated using inverse kinematics, and the residual reduction algorithm was used to improve dynamic consistency by making small changes to the torso mass center and kinematics. Muscle activations and forces were estimated from static optimization, which minimizes the sum of squared muscle activations. Knee joint reaction loads on the tibia were determined using joint reaction analysis. Data from 1-5 gait cycles were analyzed for each limb depending on the data available. Peak axial compressive and shear loads normalized to body weight were compared between children with spina bifida and controls using 2-sided t-tests with significance at $\alpha < 0.05$.

RESULTS: Knee joint reaction loads showed good consistency between trials in both controls and children with spina bifida (Figure 1). Both compressive and shear forces were lower in the spina bifida group compared with the control group ($p<0.03$) (Figure 2). This was likely due to slower walking speed in the spina bifida group (1.1 ± 0.1 vs. 1.3 ± 0.1 m/s, $p<0.0001$).

Figure 1: Consistency of joint reaction loads among multiple trials

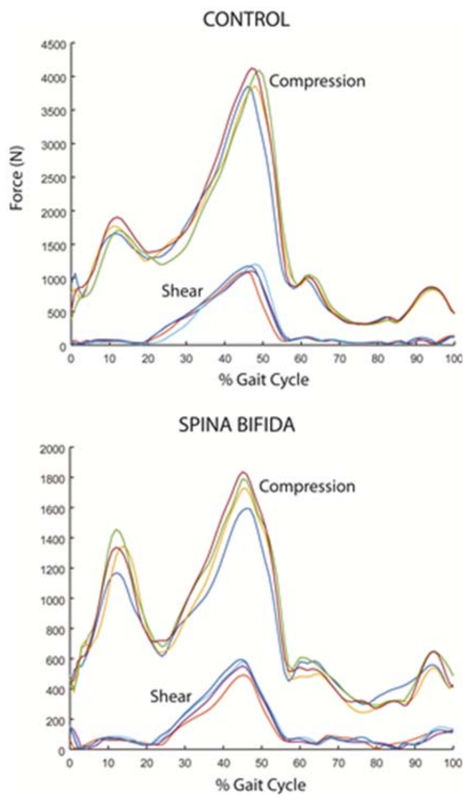
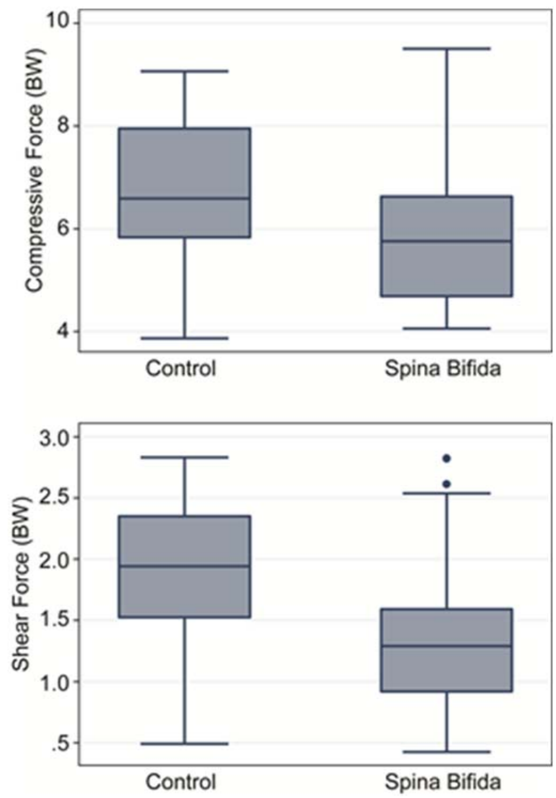


Figure 2: Comparison of joint reaction loads between control and spina bifida groups



DISCUSSION: Based on the current analysis, it appears that knee joint loads tend to be reduced, rather than elevated, in children with spina bifida. While this might lessen the risk of developing premature osteoarthritis due to overloading, the reduced loading may also contribute to lower bone mass and increased susceptibility to fractures. These findings should be considered preliminary, however, since the current analysis has several limitations. First, the model used does not have an independent degree of freedom for knee varus/valgus (this motion is coupled with knee flexion) and therefore cannot fully model the valgus thrust often observed in children with spina bifida. Also, with the current model we can only examine the resultant joint load and not the distribution of loading, which might fall disproportionately on the lateral condyle. Also, loads are higher than measured in vivo, though the effect should affect both groups similarly allowing valid comparison between groups. Finally, muscle weakness was not explicitly modeled, and muscle forces and activations have not yet been examined in detail. However, preliminary analysis indicates that even without reducing optimal muscle forces, estimated gastroc-soleus activation is much lower than normal in the children with spina bifida, as would be expected, increasing confidence in the simulation results. Despite these limitations, this study provides new insight into joint loading during gait in ambulatory children with spina bifida, which cannot be directly measured.

REFERENCES: [1] Delp et al., IEEE Trans Biomed Eng 54, 1940-50, 2007. [2] Rajagopal et al., IEEE Trans Biomed Eng 63, 2068-79, 2016.

DISCLOSURE STATEMENT: The authors have no conflicts of interest to declare.

Utilization of the Gait Profile Score (GPS) and the Gait Variable Scores (GVS) to quantify changes in gait with disease progression in boys with Duchenne Muscular Dystrophy (DMD)

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INTRODUCTION

DMD is an X-linked recessive disease of muscle characterized by a progressive loss of functional muscle mass. Historically, boys with DMD lose the ability to walk between the ages of 8-12 years, due to progressive muscle weakness. Traditionally, gait patterns have been divided into three stages, early, transitional, and late based on significant gait variables. In the early stages when boys are between 4-6 years of age, gait changes are subtle consisting of slightly increased knee and hip flexion in swing phase, as a compensatory mechanism to clear the foot in swing.[1; 2] In the transitional stage, between 6-10 years of age, there is exaggerated anterior pelvic tilt, loss of the initial knee flexion wave with weight acceptance,[1; 2] and increased foot drop in swing. In the late stages, there is a further increase in anterior pelvic tilt, widened based of support and an early heel off posture.[1] There are few published natural history studies quantifying the changes in the gait pattern since corticosteroid intervention has become a standard of care. While changes in specific kinematic patterns provide a general understanding of the gait deviations that occur with disease progression, it is difficult to quantify the magnitude of pathology with age. In recent years, indices of gait pathology have been developed to quantify the gait deviations using the information from the gait kinematics. The Gait Profile Score (GPS) is an index of pathology based on nine key gait variable scores (GVS) determined by the root mean square difference between a specific time normalized gait variable and the mean data from the reference population calculated across the gait cycle.[3] The purpose of this study was to quantify the impact of disease progression on the gait kinematics of boys with DMD using the GPS and GVS represented by the Movement Analysis Profile (MAP). We hypothesized that overall GPS and these GVS would increase (worsen) with age.

CLINICAL SIGNIFICANCE

Longitudinal MAP's specific to boys with DMD can be used to determine the magnitude of gait pathology with disease progression and gauge whether novel therapeutics alter the course of disease progression.

METHODS

In a multi-center longitudinal study, 90 ambulatory boys with DMD, ages 4-16 years at enrollment, were recruited from the MDA clinics at each of the three participating hospitals. Joint kinematics were collected using either a VICON 612 or a Motion Analysis Corporation 3-D system. All testing was performed at the child's self-selected velocity. Data were processed and analyzed at one hospital using Plug-in Gait. Boys, 4 -13.5 years of age were divided into 19 age groups based on 6-month intervals. Average files from 74 boys were used in the development of the age group MAPs, with each subject contributing a median of 3.5 visits. An overall GPS and individual GVSs were calculated for each age group. No significant difference was found between the right and left GVS and therefore the right side was used as the representative side. Control data for the MAP was developed from the gait data of 90 able-bodied children. Regression was used to determine whether GPS and all GVSs changed with age. Significance was set at $p < 0.05$.

RESULTS

Gait pathology was exhibited in boys with DMD at 4 years of age, as demonstrated by a GPS of 9.7°. As the GPS for the controls was 5.6° and the MCID for the GPS was 1.6°, this indicates that boys with DMD have a gait pathology that is 2.5 times normal and increases with age. Sagittal plane kinematics revealed a significant increase in pelvic tilt, hip flexion/extension, and ankle dorsiflexion/plantarflexion pathology with age. While the magnitude of knee flexion/extension pathology is greater than controls at 4 years of age, no significant increase was noted with age. Coronal plane deviations at the pelvis and hip significantly increased with age. No significant changes in internal/external rotation were seen with age.

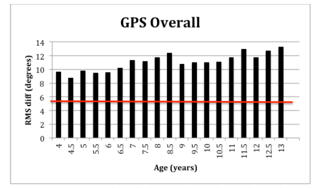

Figure 1: Representation of GPS for boys with DMD by age (Control)

DISCUSSION

Gait pathology in boys with DMD is evident at four years of age. Deviations at the pelvis, hip and ankle increased with age, while the magnitude of deviation about the knee remained constant over time. The MAP provides a method of quantifying the overall degree of gait pathology relative to controls, as well as a means of quantifying specific joint deviations in all planes of motion. These findings promote a greater understanding of the primary gait strategies used by boys with DMD to maintain ambulation as muscles weakness progresses. Due to the heterogeneity of disease progression observed in DMD, a larger sample size within each age group would improve the validity of the age group MAPs. Regardless, these data suggest that the MAP may be a useful outcome for gauging the effectiveness of novel therapeutics on the biomechanics of gait in boys with DMD.

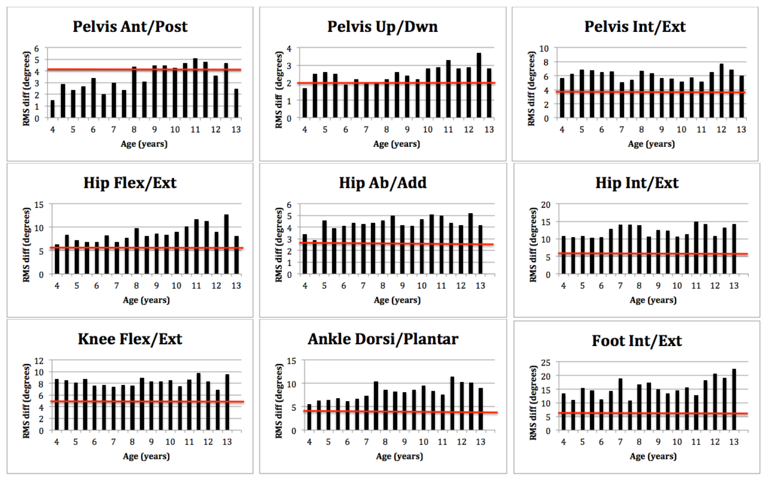

Figure 2: Representation of GVS for boys with DMD by age (Control)

REFERENCES

1. Hsu J, Furumasu J. Gait and Posture Changes in the Duchenne Muscular Dystrophy Child. Clinical Orthopaedics and Related Research 1993; 288: 122-5.

2. Sutherland D, Olsen R, Cooper L, Wyatt M, Leach J, Mubarak S, Shultz P. The Pathomechanics of Gait in Duchenne Muscular Dystrophy. Developmental Medicine Child Neurology 1981; 23: 3-22.

3. Baker R, McGinley JL, Schwartz M, Thomason P, Rodda J, Graham HK. The minimal clinically important difference for the Gait Profile Score. Gait & posture 2012; 35: 612-5.

ACKNOWLEDGMENTS

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DISCLOSURE STATEMENT

The authors have no conflicts of interest to disclose.

ANKLE DORSIFLEXION DIFFERENCES BETWEEN LUNGE TEST AND GAIT IN PATIENTS WITH CHARCOT-MARIE-TOOTH

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INTRODUCTION

Charcot-Marie-Tooth (CMT) is the most commonly inherited peripheral neuropathy with an incidence of 1 in 250,000 people [1]. Resulting gait dysfunctions are treated with a variety of conservative therapies, orthotic bracing and surgical interventions. Treatment decisions are based on clinical examination, radiographs and, when available, motion analysis. It is known that discrepancies exist between static clinical exam measures and dynamic joint function in other neuromuscular disorders [2] however, the research to understand this relationship for CMT is lacking. Previous research showed that clinical non-weight bearing ankle dorsiflexion underestimated the dynamic ankle dorsiflexion during stance in gait when ankle plantar flexor weakness was present. The focus of this abstract was to determine if a similar trend exists with a weight-bearing passive ankle dorsiflexion measure in reference to stance phase of gait.

CLINICAL SIGNIFICANCE

Understanding the relationships between clinical exam ankle measures and functional ankle range of motion during gait has important implications on treatment decision making in CMT due to progressive decline in ankle musculature strength.

METHODS

This retrospective review used a convenience sample of persons with CMT who completed instrumented 3D gait analysis following a standardized protocol. All participants performed barefoot walking at self-selected velocities. Clinical examination measures included passive ankle dorsiflexion using a lunge test and inclinometer with ankle plantar flexor strength assessed with both manual muscle test and dynamometer [3]. A clinician assessed presence of cavus deformity visually. Peak ankle dorsiflexion angle and ankle power generation in stance was extracted using custom software. Total cohort was allocated to three groups: Group 1 where dorsiflexion during gait was greater than lunge test dorsiflexion; Group 2 where gait dorsiflexion was equal to lunge test (defined as within 5°); Group 3 where gait dorsiflexion was less than lunge test range of motion. Comparisons of the group means were analyzed using Pearson correlation coefficient and chi square analysis with *p* value significance set at <0.05.

RESULTS

A total of 50 sides from 11 males, 14 females with average age of 14.4 years (SD ±4.9) were analyzed. Initial correlations of peak ankle dorsiflexion in stance to clinical exam dorsiflexion and lunge test were weak (r=0.19 and r=0.10 respectively). However correlation of peak ankle dorsiflexion in stance to plantar flexor strength on manual muscle test and by dynamometer were stronger (r=-0.6 and r=-0.66 respectively). As expected, a significant difference was present between Group 1 and Group 3 from Group 2 in peak dorsiflexion angle in stance (Table 1). There was no signficant difference in body mass index or age between groups (Table 1). There were significant differences between Group 1 and 2 in ankle plantar flexor strength (0.0014), peak ankle power generation in stance (p=0.0017) and frequency of cavus deformity (0.019). Interestingly there was no significant difference in plantar flexor strength or peak ankle power generation between Groups 2 and 3 despite a difference in median plantar flexor strength grade on manual muscle test.

Table 1. Group values represented as mean ±SD except for Plantar flexor Strength (MMT) represented as the median. **p* value significant at <0.05

	Group 1 (n=7)	**Group 2 (n=29)**	**Group 3 (n=12)**	***p* value** Group 1 vs Group 2	***p* value** Group 2 vs Group 3
Age (years)	15.6 ± 8.02	14.6 ± 4.7	13.8 ± 3.6	0.756	0.556
Body Mass Index (kg/m2)	21.0 ± 3.74	22.9 ± 5.4	23.1 ± 3.3	0.295	0.527
Lunge Test Dorsiflexion (°)	14.5 ± 9.0	16.5 ± 5.9	21.4 ± 5.4	0.055	0.018*
Peak Dorsiflexion Angle in Stance (°)	22.4 ± 5.2	17.2 ± 4.5	11.9 ± 5.9	0.040*	0.013*
Plantar flexor Strength (MMT)	2	4	5	na	na
Plantar flexor Strength (N)	25.3 ± 26.1	106.5 ±44.5	153.9 ± 41.4	0.0014*	0.07
Frequency of Cavus deformity (%)	0%	48%	33%	0.019*	0.781
Peak Ankle Power Generation in Stance (W/kg)	1.4	2.4	3.2	0.0017*	0.069

DISCUSSION

In the presence of ankle plantar flexor weakness, a weight-bearing ankle dorsiflexion measure underestimates the peak ankle dorsiflexion that occurs during stance in CMT. While Group 3 showed increased passive lunge dorsiflexion range of motion compared to gait, they likely have the ability to control the forward tibial progression in stance within their available range of motion given adequate plantar flexor strength. The signficant plantar flexor weakness in Group 1 contributes to an inability to prevent hyperdorsiflexion in stance that was not appreciated with their passive lunge test alone. The sole use of passive ankle dorsiflexion measures, although weight bearing, is inadequate to predict dynamic ankle function during gait in the presence of ankle plantar flexor weakness and the absence of motion analysis. Further understanding of a predictive relationship should be examined.

REFERENCES

1. Martyn CN et al. (1992) Journal of Neurology, Neurosurgery and Psychiatry, 62:310-318.
2. McMulkin M. et al. (2000) Journal of Pediatric Orthopedics, 20:366-369.
3. Burns J, et al., (2012) Annals of Neurology 71(5), 642-652.

DISCLOSURE STATEMENT

The authors have no conflicts of interest to disclose.

RECTUS SPARING APPROACH TO PERIACETABULAR OSTEOTOMY IN ADOLESCENTS PRESERVES HIP FLEXION STRENGTH AT SHORT TERM FOLLOW-UP

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INTRODUCTION

The Bernese periacetabular osteotomy (PAO) [1] has proven to be a reliable surgical technique for treating symptomatic acetabular dysplasia in skeletally mature patients, resulting in significant decreases in pain, improved function post-operatively and few reported complications. The original description of the PAO includes release of the direct and indirect heads of the rectus femoris tendon for complete visualization of the anterior hip capsule (with repair of the rectus tendon through bone at the end of the procedure).[2] We have observed prolonged rehabilitation and delayed return to activities associated with slow return of rectus strength, with a similar trend in hip flexor pull-off power[3]. Presented with these results, we therefore adopted a rectus-sparing approach to the PAO, with no release of the rectus tendons.

Recent studies have demonstrated that a rectus sparing approach to the PAO is safe and as effective as the original approach in achieving appropriate orientation of the acetabulum.[4-5] The purpose of this study was to compare: 1) hip flexion strength 2) hip flexion pull-off power during gait and 3) self-reported patient outcomes scores at 6mos and 1 yr post-operatively in patients who had undergone a PAO with the traditional approach or a PAO with a rectus sparing approach (PAO-RSA).

CLINICAL SIGNIFICANCE

Preserving the rectus femoris when performing a PAO may lead to improved short term conservation of hip flexor strength with greater return to activities in the early post-operative period and lead to improved health related quality of life.

METHODS

This was an IRB approved consecutive series of adolescents with radiographic evidence of dysplasia, no underlying neuromuscular disease, with a planned treatment of a Bernese PAO who completed pre-op and post-op gait analysis (6mo and 1yr). Gait analysis included hip kinematics and kinetics (Vicon Nexus). Isokinetic hip flexion and abductor strength (Biodex System 3, 60°/sec) were determined using peak torque normalized to body weight.

Standard radiographic measurements, including the lateral center edge angle (LCEA) acetabular index of the weight bearing zone (AI) and the ventral center edge angle (VCEA), were analyzed at pre-op and 1 year post-op to quantify the initial acetabular dysplasia and the correction achieved. Patients were asked to complete the modified Harris Hip Score (mHHS). Differences between the radiographic measures, gait parameters and the functional hip scores at the pre-operative and follow-up evaluations were analyzed using a paired t-test.

RESULTS

The **PAO-RSA** group (10 hips [8 females], average age at surgery 16±1yrs) were compared to a subset of 13 patients who were matched for pre-operative hip strength from a previously reported cohort of PAO patients (**PAO**)]3].

Radiographically, there were no differences in pre-op deformity, post-op correction or degree of correction between the PAO and PAO-RSA groups. Hip flexor strength and pull-off power were decreased 6 months in the PAO group compared to the PAO-RSA group, but showed delayed improvement leading to no significant differences in strength at 1 year. There were no differences between groups in mHHS any time point and both groups improved significantly post-postoperatively.

Table 1. The PAO-RSA was able to preserve hip flexion strength, as well as hip flexion pull-off power at 6mo post-op compared to the PAO group

		PAO-RSA	PAO	p-value
Hip Flexion Strength (%Nm/kg)	**Pre**	102.3 ± 33.1	100.0 ± 26.3	ns
	6mo post	95.8 ± 42.9	65.3 ± 21.9	0.037
	1yr post	90.1 ± 41.2	79.6 ± 32.1	ns
Hip Flexion Pull-Off Power (W/kg)	**Pre**	1.39 ± 0.40	1.63 ± 0.60	ns
	6mo post	1.76 ± 0.44	1.33 ± 0.29	0.01
	1yr post	1.91 ± 0.98	1.45 ± 0.49	ns
mHHS (max 89)	**Pre**	67 ± 9	63 ± 11	ns
	6mo post	79 ± 8	71 ± 12	ns
	1yr post	75 ± 12	79 ± 8	ns

DISCUSSION

The rectus sparing approach is safe and allows for unimpeded re-orientation of the acetabulum [4-5] but there has been no previous study evaluating hip flexion strength when using the approach. In this small cohort, it was demonstrated that the rectus-sparing approach to the PAO preserves hip flexion strength in adolescents with hip dysplasia at 6 months post-operatively when compared to pre-operatively and when compared to those who underwent the traditional approach. Release of the indirect and direct heads of the rectus femoris tendon, as originally described, results in decreased hip flexion strength at 6 months and 1 year post-operatively when compared to pre-operative strength.

Although this report is limited to 1 year follow-up, it is during this immediate post-operative period that the majority of the strength limitations/improvements are seen. Clinically, this study mirrors previous results in terms of radiographic correction, acetabular fragment mobilization/final positioning and self-reported function (HHS) in both groups.

In conclusion, preserving the rectus femoris when performing a PAO leads to improved short term conservation of hip flexor strength and hip flexion pull off power. This may lead to earlier return to activities in the early post-operative period and may provide for improved health related quality of life and activity scores in the future. Further assessment at long-term follow-up is needed to determine if this strength leads to further improvement in functional outcomes.

REFERENCES: 1. Ganz et al. *Clin Orthop*1988;232:26-36. **2.** Beaule PE *CORR.* 2015;473:615-622. **3.** Sucato DJ et al. *JPO.* 2010;30:344-350. **4.** Novais EN et al. *CORR.* 2014. **5.** Peters CL et al. *CORR.* 2015;473:608-614.

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DOES HIP ABDUCTOR FUNCTION INCREASE AFTER A FEMORAL DEROTATION OSTEOTOMY IN INDIVIDUALS WITH CEREBRAL PALSY?

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INTRODUCTION

One reason to perform a femoral derotation osteotomy (FDO) in children with cerebral palsy (CP) is to correct coronal plane hip abductor moment arms, thus functionally strengthening the abductors. Despite this widespread logic, there is a paucity of hip kinetics reported post-FDO [1,2]. We measured if hip abductor moment, a functional measure of hip abductor utilization, increased approximately one year and five years post-FDO in individuals with bilateral CP.

CLINICAL SIGNIFICANCE

Hip abductor moment does not improve in the short-term post-FDO but does improve after implant removal three or more years later. This information should be used to counsel patients.

METHODS

A retrospective analysis of our database was performed to identify individuals with CP who were ≤18 years old at preoperative analysis and could ambulate without an assistive device. Additional gait analyses that were approximately one year (*short-term*) and at least three years post-FDO (*mid-term*) were compared. The primary outcome was dimensionless mean hip abductor moment during single-support [3]. Secondary outcomes included anteversion, mean hip rotation, and estimated coronal plane hip abductor moment arm derived from a musculoskeletal model that included both anteversion and hip rotation [4]. Wilcoxon signed rank or paired t-tests were used as appropriate ($p<0.05$). A Bonferroni correction was applied.

RESULTS

There were 135 individuals who met the short-term criteria and 37 for mid-term. Most FDOs were performed proximally. Hip abductor moment remained unchanged at short-term (Table 1). This was unexpected given that anteversion and internal hip rotation decreased 35° and 13°, respectively, which increased estimated hip abductor moment arm. On average, hip abductor moment at mid-term improved from 43% less than to 23% less than typically developing controls. For 49% of individuals, the increase in moment (pre- to mid-term) exceeded our minimal detectable change, while it did not change for 43%, and decreased for 8%. The overall improvement in moment was observed along with 11° of recurrent internal hip rotation (which would increase moment arms) and no change in other kinematics, strength, or walking velocity.

DISCUSSION

The unchanged hip abductor moment from pre- to short-term may be attributed to gait compensation that unloads the hip to avoid pain from retained implants, whereas the increase from short- to mid-term may be due to implant removal or greater rehabilitation time. The 5 year post-FDO increase in hip abduction moment underscores the benefits of an FDO into adolescence for independent ambulating individuals with CP.

Table 1. Individual characteristics and outcomes (median(IQR)) from pre- to short- for 135 individuals (top half) and from pre- to short- to mid-term gait analyses for 37 individuals (bottom half).

	Age (yrs) [range]	**Mean hip abductor moment in single-support** (non-dimensional)	**Mean hip rotation in single-support** (°)	**Mean pelvic obliquity in single-support** (°)	**Mean trunk obliquity in single-support** (°)	**Walking velocity** (non-dimensional)	**Anteversion** (°)	**% change in abductor moment arm**	**Isometric hip abductor strength**
SHORT-TERM, n=135									
Pre-operative	9.4(3.9) [3.7-17.2]	0.033(0.027)	11.3(12.5)	1.4(4.9)	-3.2(7.1)	0.39(0.10)	50(15)	-19(21)	3(0)
Short-term	10.9(4.0) [5.4-18.3]	0.032(0.029)	-2.1(12.4)*	0.6(5.0)	-4.1(6.5)	0.37(0.09)*	15(10)*	2(12)*	3(0)
p-value		0.802	<0.001	0.207	0.900	0.048	<0.001	<0.001	0.723
MID-TERM, n=37									
Pre-operative	8.8(4.3) [4.1-15.1]	0.032(0.034)	10.7(11.9)	1.2(5.5)	-4.8(8.4)	0.36(0.10)	50(16)	-19(29)	3(0)
Short-term	10.7(4.9) [5.4-16.8]	0.035(0.026)	-3.2(11.6)*	1.3(5.2)	-5.1(6.0)	0.37(0.09)	15(10)*	1(11)*	3(0)
Mid-term	14.4(3.9) [7.6-18.3]	0.043(0.032)^~	7.8(19.9)^~	2.8(4.2)	-4.6(5.2)	0.34(0.09)	15(10)^	6(10)^	3(2)
Typically developing (mean±SD)		0.056±0.015	1.6±8.9	1.3±2.0	-0.1±3.2	0.43±0.0.07	19		5

Positive values indicate hip abductor moment, internal hip rotation, contralateral pelvic drop, contralateral trunk lean, and abductor moment arm above normal.
* significant change (p<0.05) from pre- to short-term
^ significant change (p<0.05) from pre- to mid-term
~ significant change (p<0.05) from short- to mid-term

REFERENCES

1. Cimolin, V. et al. (2011) Hip International, 21(6):657-664.
2. Niklasch, M. et al. (2015) Gait & Posture, 42(4):460-465.
3. Pinzone, O. et al. (2015) Gait & Posture, 44:68-73.
4. Arnold, A. et al. (1997) Developmental Medicine & Child Neurology, 39(1):40-44.

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Excessive Femoral Anteversion Does Not Cause Excessive Anterior Pelvic Tilt

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INTRODUCTION

Excessive femoral anteversion is a common finding among children with cerebral palsy (CP). It has been suggested that excessive femoral anteversion contributes to excessive anterior pelvic tilt, also commonly observed in children with CP [1-4]. The mechanism for this effect is generally described as follows: 1) a person adopts an internally rotated hip position to compensate for excessive femoral anteversion, 2) in this position, the lesser trochanter moves posteriorly, 3) as a consequence of the posterior lesser trochanter, the psoas is stretched, 4) the tension in the psoas results in anterior tilting of the pelvis through coupling with the lumbar spine. Additionally, increased pelvic tilt is proposed as a mechanism to provide coverage of the anterior portion of the femoral head [3].

CLINICAL SIGNIFICANCE

Excessive femoral anteversion does not cause anterior pelvic tilt. As a result, the rationale for an FDO, and the choice of FDO technique (i.e. proximal vs. distal), should not include improving excessive anterior pelvic tilt.

METHODS

This study involves a retrospective analysis of a large clinical database. Inclusion criteria were a primary diagnosis of bilateral CP, availability of gait data from two routine clinical gait analysis evaluations, and no prior FDO. Linear regressions were used to examine the relationship between femoral anteversion and mean pelvic tilt for various subsets of limbs.

RESULTS

For limbs with no prior surgery ($N = 777$), there was no relationship between femoral anteversion and pelvic tilt ($p > .05$) [Figure 1A]. For limbs with no prior FDO ($N = 3193$), there was also no relationship between femoral anteversion and pelvic tilt ($p > .05$) [Figure 1B].

Limbs undergoing single-event multi-level surgery (SEMLS) with no previous FDO were split into those where the SEMLS included an FDO (SEMLS+FDO) or did not include an FDO (SEMLS-FDO). There was no relationship between preoperative femoral anteversion and pelvic tilt in either group (SEMLS+FDO: $N = 862$, $p > .05$, SEMLS-FDO: $N = 213$, $p > .05$) [Figure 1C]. We further examined changes in pelvic tilt following SEMLS $\pm$FDO and found no relationship between change in anteversion and change in pelvic tilt ($p > .05$ for both +FDO and –FDO groups) [Figure 1D].

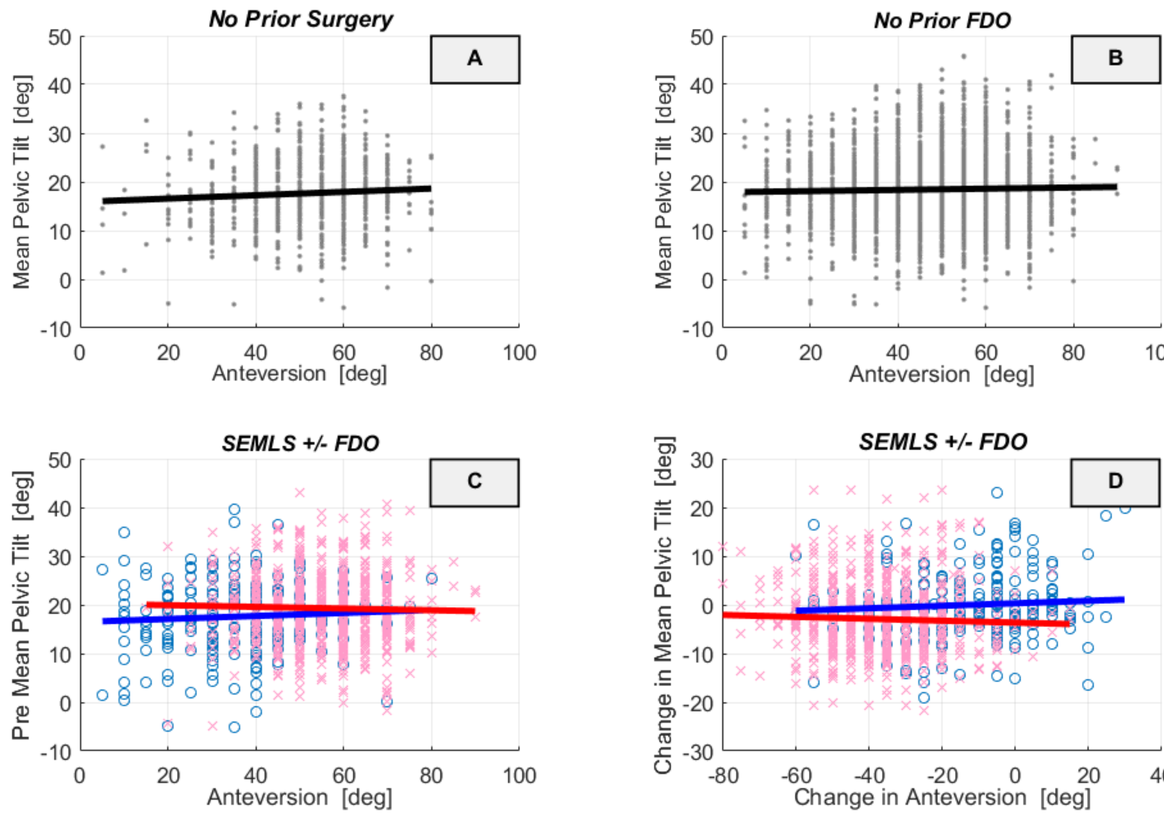

Figure 1. [A] There was no relationship between anteversion and pelvic tilt for limbs of individuals with CP who had no prior surgery, **[B]**there was no relationship between anteversion and pelvic tilt for limbs of individuals with CP who had no prior FDO, **[C]** there was no relationship between preoperative anteversion and preoperative pelvic tilt for limbs undergoing SEMLS with no prior FDO, with (**red**) or without (**blue**) an FDO as part of the SEMLS, and **[D]** there was no relationship between ***change*** in anteversion and ***change*** in pelvic tilt for limbs undergoing SEMLS with no prior FDO, with (**red**) or without (**blue**) an FDO as part of the SEMLS.

DISCUSSION

There is no relationship between anteversion and pelvic tilt. There is also no relationship between change in anteversion and change in pelvic tilt following SEMLS. This is true regardless of whether or not the change in anteversion is the result of an FDO. Together, these findings strongly suggest that excessive femoral anteversion *does not cause* excessive anterior pelvic tilt. This finding raises important questions about surgical technique, since the proximal approach to an FDO is, in part, motivated by the chance to “unwind” the lesser trochanter as a means of slackening the psoas and improving anterior pelvic tilt.

REFERENCES

1. Akalan E et al. (2013) *Hip Int*, 23(5):492-9.
2. Desloovere K (2008) *Gait Posture*, 28S:7-8.
3. Gage JR et al. editor (2009) *The Identification and Treatment of Gait Problems in Cerebral Palsy*, Mac Keith Press, 194.
4. Gage JR et al. (2013) *Dev Med Child Neurol*, 55(S3): 21.

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THE LONG-TERM OUTCOME OF PELVIC ASYMMETRY DURING GAIT IN CHILDREN WITH CEREBRAL FOLLOWING UNILATERAL FEMORAL DEROTATION OSTEOTOMY

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INTRODUCTION: Cerebral palsy (CP) is a motor disorder caused by a static, non-progressive lesion in the immature brain. Gait deviations are common in children with CP including internal hip rotation and pelvic asymmetry [1,2,3]. Femoral derotation osteotomy (FDO) is a surgical procedure aimed to correct the internal torsion of the femur which has been shown to be effective in the short-term with slight recurrence long-term [4,5]. However, little attention and research has focused on the pelvic asymmetry outcome. This study aims to determine the change in pelvic rotation in the short and long-term following unilateral femoral derotation osteotomy.

CLINICAL SIGNIFICANCE: Clinical significance resides in surgical technique influence. The goal is to determine the long-term outcome of femoral derotation osteotomy on children with internal hip rotation and pelvic asymmetry. Determination could potentially influence the physician's surgical technique to achieve long-term maintenance of the correction torsional alignment.

METHODS: In this IRB-approved retrospective study, children with CP who underwent a unilateral femoral derotation were included that had a preoperative gait analysis, as well as a short term (ST; 1-3 years postoperative) and/or had a long term (LT; >5 years postoperative) gait analysis. Gait analyses were completed in the motion analysis lab using 3D motion capture to analyze lower extremity movement during walking. Children were categorized at preoperative analysis as retracted (<-4 degrees), neutral (-4 to +4 degrees) or protracted (>+4 degrees) pelvic position on the operative side. Neutral values were based on the mean ± 1SD of lab normative data. The children were placed in two groups by GMFCS level: GMFCS I/II and GMFCS III/IV. Statistical analyses using ANOVA and Chi-Square were completed.

RESULTS: In 46 children (age 10 ± 4 years at surgery), at pre-operative analysis 70% had pelvic retraction, at ST (2 ± 1 years post) only 40% of children had pelvic retraction, but at LT (8 ± 2 years post) 64% of children recorded pelvic retraction on the operative side (Table 1). While a trend is present in the frequency analysis of the ST and LT assessments, chi-square test shows that it is not statistically significant (p = 0.078). Pelvic retraction at pre-op was -20° ± 10° and improved to -6° ± 12° at ST ($p<0.00117$, pre-op vs ST), and was maintained at LT (-8° ± 13°; $p<0.00938$, pre-op vs LT; Table 2; n=15). The GMFCS I/II group improved pelvic symmetry and hip rotation at ST ($p<0.01$), with a trend to maintain correction at LT ($p<0.1$; pre-op vs LT). GMFCS III/IV demonstrated a similar average change in pelvic symmetry and hip IR at ST but due to greater variation in ST outcome the change was not significant (Pelvis p=0.22, Hip p=0.05). Greater recurrence in both pelvic retraction and hip IR were present in the GMFCS III/IV group with no significant difference between pre-operative and LT (Pelvis p=0.73; Hip p=0.48).

Table 1: Percentage of the total number of children with retracted, neutral, or protracted pelvis from pre-op, ST and LT gait analyses.

	n	Retracted (%)	Neutral (%)	Protracted (%)
Preoperative	46	70	13	17
Short term	40	40	23	37
Long term	25	64	16	20

Table 2: Pelvic and hip rotation before (pre-operative) and after (post-operative) unilateral femoral derotation osteotomy. ST: short term (1-3 years follow up); LT: long term (≥5 years follow up); Pelvis: pelvic rotation mean in stance (- retracted); Hip: hip rotation mean in stance (+ internal).

Angles (deg)					p-values		
	PreOp	ST	LT	Normal	Pre-op & ST	Pre-op & LT	ST & LT
Pelvis	-20 (10)	-6 (12)	-8 (13)	0 (4)	0.00117	0.00938	0.585
Hip	17 (8)	6 (13)	11 (15)	-6 (8)	0.0065	0.198	0.279

DISCUSSION: In those children with pelvic retraction related to asymmetric hip IR, unilateral FDO corrects retraction in the short term, but with recurrence of hip IR at LT, pelvic retraction also recurs. This data suggests a strong link between the hip and pelvis. With greater recurrence seen in those children classified as either GMFCS III or IV, the effect of functional level on recurrence should also be considered. Clinically, these results could influence surgical technique in order to maintain correction of pelvic asymmetry in the long term.

REFERENCES

1. Dreher, T et al. (2007) Gait Posture. 6;26(1):25–31.
2. Miller, F. (2005) Cerebral Palsy. Springer Science & Business Media; 1055 p.
3. O’Sullivan, R et al. (2006) J Pediatr Orthop. Jul;26(4):537–41
4. Dreher, T. et al. (2012) Gait Posture. Jul;36(3):467–70.
5. Kim, H. et al. (2005) J Pediatr Orthop. Nov;25(6):739–43.

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Podium Session #6

TRUNK AND UPPER EXTREMITY

MODERATED BY: **Joanna Roybal, PT:** Center for Gait and Movement Analysis
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1. **Comparison of Temporal Spatial Characteristics and Shoulder Kinematics and Kinetics During Reverse and Conventional Manual Wheelchair Propulsion in Persons with Paraplegia**
   Lisa Lighthall Haubert, Philip Requejo, Sara Mulory, Somboon Maneekobkunwong, Diego Rodriguez, JoAnne Gronley

2. **Scapular Kinematics in Adolescents with Idiopathic Scoliosis**
   Elizabeth Rapp, R. Tyler Richardson, Ross Chafetz, Amer Samdani, James Richards

3. **Comparing Measurements of Shoulder Motion by A Physician, by an Occupational Therapist with a Goniomenter and by Motion Capture**
   Stephanie Russo, Ross Chafetz, Luisa Rodriquez, Carolyn Roposh, Dan Zlotolow, Scott Kozin, James Richards

4. **Evaluation of Shoulder Stretching with and Without Scapular Stabilization in Children with Brachial Plexus Injuries**
   Stephanie Russo, Carolyn Killelea, Luisa Rodriguez, Ross Chafetz, Scott Kozin, Dan Zlotolow, James Richards

5. **Head Kinematics and Head-trunk Coordination in People With and Without Vestibular Hypofunction**
   Serene Paul, Raymond Walther, Mark Lester, Clough Shelton, Richard Gurgel, Lee Dibble

6. **Effect of Cervical Decompression Surgery on Gait in Adult Cervical Spondylotic Myelopathy Patients**
   Ram Haddas, Kevin Ju, Raj Arakal, Theodore Belanger

7. **A Comprehensive Method to Measure Three-dimensional Breast Motion During Physical Activity**
   Sarah Colon, Elisa Arch, James Richards

8. **Effect of Load Carriage on Lower Extremity Coordination Variability During Constant Speed Treadmill Marching in ROTC Cadets**
   Braden Romer, Scott Stetson, David Szymanski, Meagan Arflin, Hocheng Lu

COMPARISON OF TEMPORAL SPATIAL CHARACTERISTICS AND SHOULDER KINEMATICS AND KINETICS DURING REVERSE AND CONVENTIONAL MANUAL WHEELCHAIR PROPULSION IN PERSONS WITH PARAPLEGIA

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INTRODUCTION

Shoulder pain after spinal cord injury (SCI) is attributed to a shift in mobility demands to the upper limbs and negatively impacts independence, participation and quality of life[1, 2].

CLINICAL SIGNIFICANCE

Repetitive superior and posterior shoulder joint forces produced during wheelchair (WC) propulsion can result in impingement of subacromial structures if unopposed, owing to fatigue or weakness[3]. Further, these forces increase with fast and inclined propulsion. RoWheels® (RW), geared rear wheels that produce forward WC movement with backward rim pulling, have the potential to reverse these shoulder joint forces and utilize the larger posterior shoulder and scapular muscles to protect the shoulder joint and preserve mobility.

METHODS

Ten men with paraplegia from SCI (AIS A, B) who pushed traditional manual WCs and were free of shoulder pain (Wheelchair User's Shoulder Pain Index (WUSPI) < 12) or pathology participated. Right upper extremity/trunk kinematics and kinetics were collected during three conditions of ergometer propulsion[4] sitting on their own seat cushion in a rigid, lightweight WC frame adjusted to match the participant's own WC: self-selected free speed reverse propulsion (pulling back on the rim) with RW, and matched-speed reverse (rSW) and forward (fSW) propulsion with standard instrumented pushrim wheels (Smartwheels (SW) using an analog display of propulsion speed. Right 3-D pushrim kinetics were collected on the right at 200 Hz using SW. Three-dimensional trunk, right upper extremity, and wheel kinematics were collected at 200 Hz using a Qualisys motion capture system (10 cameras). Kinematic and kinetic data were low-pass filtered with a fourth-order zero-lag Butterworth filter with cutoff frequencies of 8 Hz and 10 Hz, respectively, using Visual3D (C-Motion, Inc., Germantown, MD).

Temporal-spatial, kinematics and kinetics variables were compared across the three conditions (RW, fSW and rSW) with a repeated-measures Analysis of Variance.

RESULTS

Mean age of participants was 39.6 years and duration of SCI was 14.5 years. Free propulsion velocity and cadence were similar during RW and rSW compared to fSW, although push distance was mildly reduced in rSW vs. RW and fSW (Table 1).

At the forward-most hand contact position (initial contact during RW and rSW, and end contact during fSW), shoulder flexion, abduction and internal rotation were similar in all three conditions. At the backward-most hand position (end contact during RW and rSW and beginning contact during fSW propulsion) shoulder extension was significantly lower in both RW and rSW vs. fSW propulsion. Shoulder abduction and internal rotation in the backward-most hand position was similar in all three conditions.

Anteriorly and inferiorly directed forces on the pushrim were significantly greater during fSW than in rSW propulsion. In contrast, posteriorly and superiorly directed forces on the pushrim were significantly greater during rSW than in fSW. Medial and lateral forces on the pushrim were similar between the two conditions.

Superior (upward) force at the shoulder joint was significantly higher during fSW than during rSW propulsion. Posterior shoulder force was also greater during fSW than rSW. Inferior force tended to be higher during rSW than fSW. Anterior and medial shoulder forces were similar between fSW and rSW conditions.

Peak flexor, adductor, and external rotation moments during the contact phase (forward or reverse) were significantly higher during fSW propulsion than during rSW. Shoulder moments during the contact phase were otherwise similar during fSW and rSW propulsion.

Table 1. Temporal-spatial, kinematics and kinetics data for RW, rSW and fSW.

	RW	**rSW**	**fSW**
Velocity (meters/minute)	70.1 ± 10.9	70.3 ± 6.9	72.0 ± 7.8
Cadence (pushes/minute)	67.4 ± 26.1	68.1 ± 14.1	61.5 ± 17.1
Push Distance (meters)	1.16 ± 0.35	*1.06 ± 0.17	1.23 ± 0.27
Shldr EXT (°)**, backward hand position**	25.5 ± 10.3	25.6 ± 16.9	*41.2 ± 9.9
Pushrim Forces (Newtons(N))**, Anterior**	-	5.2 ± 7.8	*33.6 ± 4.3
Pushrim Forces (N)**, Inferior**	-	25.0 ±16.6	*44.6 ± 11.7
Pushrim Forces (N)**, Posterior**	-	*35.9 ± 13.8	3.3 ± 4.3
Pushrim Forces (N)**, Superior**	-	*22.5 ± 21.8	0.6 ±0.7
Shldr Forces (N)**, Superior**	-	-3.2 ± 15.4	*12.7 ± 10.7
Shldr Forces (N)**, Posterior**	-	13.0 ± 8.9	*41.7 ± 7.7
Shldr Forces (N)**, Inferior**	-	**56.7 ± 7.7	48.4 ± 10.2
Shldr Flexion Moment (Newton·meters/kg (Nm/kg))	-	5.0 ± 3.9	*14.1 ± 2.4
Shldr Adduction Moment (Nm/kg)	-	4.5 ± 2.0	*9.8 ± 2.5
Shldr External Rotation Moment (Nm/kg)	-	3.1 ± 1.8	*8.7 ± 2.3

*$p<0.05$; **$0.05>p>0.10$

DISCUSSION

These results demonstrate that reverse propulsion requires an upward force applied to the pushrim which, combined with the weight of the arm, translates to a distraction force at the shoulder. In contrast, forward propulsion incorporates a downward pushrim force leading to a superior, potentially impinging shoulder joint force. Moreover, the overall demands on shoulder muscle groups are lower during the contact phase of pulling compared to forward pushing. Reverse propulsion may protect the subacromial structures and thereby prevent injury and pain and preserve mobility, independence, and participation for individuals living with paraplegia.

REFERENCES

1. Kemp et al. (2011). J Spinal Cord Med. 34(3): 278–284.
2. Ataoğlu et al. (2013). Spinal Cord. 51(1):23–6.
3. Requejo, PS et al. (2015) Topics in Geriatr Rehabil. 31(1): 26–41.
4. Requejo, PS et al. (2015) Topics in SCI Rehab. 21 (4): 313-324.

DISCLOSURE STATEMENT

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SCAPULAR KINEMATICS IN ADOLESCENTS WITH IDIOPATHIC SCOLIOSIS

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INTRODUCTION

Adolescent idiopathic scoliosis (AIS) involves an asymmetry in the trunk and upper extremity [1]. Other trunk deformities (e.g. hyperkyphosis) have been associated with scapular dyskinesis [2], suggesting that shoulder motion may also be altered in AIS. Abnormal scapulothoracic (ST) motion, such as deficits in posterior tilting and upward rotation can lead to glenohumeral pathology [3].

The single previous study examining ST kinematics in AIS revealed more anterior tilt on the convex side and less upward rotation on the concave side during rest but not during motion [4]. However, this study only evaluated adolescents with moderate scoliosis (Avg. Cobb angle: 34°) and only evaluated ST motion during single plane humeral elevation.

The objective of this study was to evaluate ST kinematics in AIS during multiplanar arm motion, as well as examine the relationship of altered kinematics to curve severity. We hypothesized that adolescents with idiopathic scoliosis would demonstrate different ST kinematics than their typically developing peers and that these differences would be exacerbated with increased curve severity.

CLINICAL SIGNIFICANCE

This study evaluated ST function and its relationship to curve severity in AIS, providing direction for clinical consideration of the upper extremity implications of scoliosis.

METHODS

Eighteen typically developing (TD) adolescents were recruited (Mean age: 14.9 ± 1.8y) as well as 14 adolescents with a right thoracic scoliotic curvature (Mean age: 14.8 ± 1.9y). Cobb angles were obtained from the most recent radiograph of the AIS group. Markers were placed on the trunk (T1, T8, sternal notch), humerus (medial/lateral humeral epicondyles), and the acromion process. Subjects sat on a stool and held each of four positions: neutral, abduction, external rotation, and flexion. In each position, scapular landmarks (trigonum spinae and inferior angle) were palpated and markers were affixed. Marker position was recorded by a 10 camera optoelectronic motion capture system operating at 60 Hz. Coordinate systems were constructed according to ISB recommendations [5], and ST angles were calculated using a YXZ Euler sequence. A mixed ANOVA was employed to compare ST angles between TD and scoliosis adolescents in all four positions and along each axis of ST motion. Right and left shoulders were compared separately. For positions/axes with significant differences, Pearson correlations were calculated between Cobb angles and ST angles for all subjects.

RESULTS

Right shoulders (convex side) in the AIS group demonstrated significantly less posterior/more anterior tilt than TD adolescents in all positions. Left shoulders (concave side) in the AIS group demonstrated significantly more posterior tilt than TD adolescents, but only at rest (Table 1). Correlations between AIS right shoulder ST posterior tilt and Cobb angle ranged from -0.48 to -0.61. An example for the flexion position is shown in Figure 1.

Table 1: Means (SDs) of ST angles in each position. X represents upward (+)/downward (-) rotation, Y represents internal (+)/external (-) rotation and Z represents posterior (+)/anterior (-) tilt.

	Abduction			External Rotation			Flexion			Neutral (rest)		
	X	Y	Z	X	Y	Z	X	Y	Z	X	Y	Z
TD Right	47.4 (8.3)	33.7 (9.4)	-1.9 (7.4)	4.7 (9.9)	22.8 (6.0)	0.4 (4.0)	51.2 (7.1)	31.9 (7.5)	-1.6 (7.5)	3.2 (8.5)	38.5 (6.8)	-3.8 (4.1)
AIS Convex	42.8 (11.)	39.2 (8.9)	-9.3 (8.2)	-0.5 (7.4)	26.6 (6.3)	-5.2 (7.2)	42.7 (12.9)	39.9 (10.0)	-9.0 (9.2)	-4.0 (7.3)	40.7 (6.2)	-7.5 (6.9)
TD Left	44.1 (7.8)	40.1 (10.8)	0.5 (6.6)	4.6 (8.3)	22.1 (6.6)	0.4 (3.3)	46.3 (6.5)	38.1 (10.4)	2.0 (6.3)	0.7 (6.9)	38.2 (8.3)	-1.9 (4.0)
AIS Concave	41.0 (9.9)	38.4 (11.9)	1.9 (7.4)	5.4 (12.6)	23.6 (8.2)	3.7 (7.3)	44.9 (10.7)	42.4 (11.2)	-0.2 (10.8)	-0.2 (11.7)	39.4 (11.1)	3.6 (6.2)

Significant results ($p < 0.05$) indicated with shaded cell. Blue represents lower, yellow represents higher angles in the AIS group.

DISCUSSION

While the previous study only revealed posterior tilt differences at rest, these results demonstrated that abnormal scapular orientation persisted in positions across multiple planes. The discrepancy in findings may be due to choice of measurement technique, as the approach used in the Lin study produces errors at extreme levels of humeral elevation [6]. The flexion position also revealed a reduction in upward rotation and an increase in internal rotation (winging). This suggests that in order to achieve maximum humeral elevation in the sagittal plane, the scapula moves in an abnormal pattern to navigate the rib hump often present in AIS. Inadequate posterior tilting and upward rotation result in a smaller subacromial space, placing these adolescents at risk of glenohumeral pathology [4]. Additionally, the negative correlations of posterior tilt with Cobb angle suggest these deficits are exacerbated with more severe curvature. As curve progression is common in scoliosis [1], these findings represent an important consideration of increased risk for upper extremity pathology in the scoliosis population.

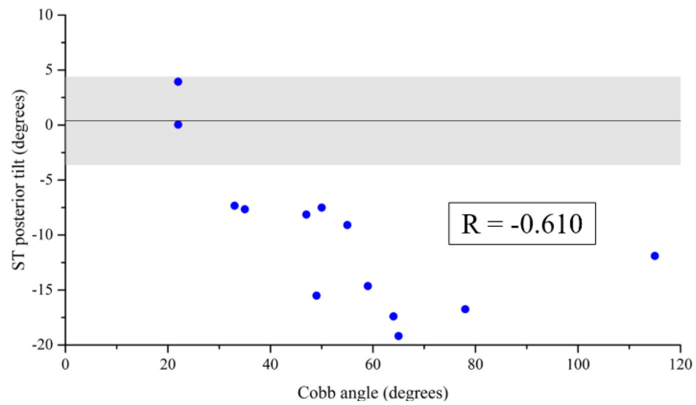

Figure 1: ST posterior tilt vs. Cobb angle. The grey bar represents mean ± 1SD of TD right scapulae in the flexion position. Blue dots represent each AIS right scapula.

REFERENCES

1. Asher, M and D. Burton. (2006) Scoliosis, 1(1):2.
2. Kebaetse, M. et al. (1999) Archives of Physical Medicine and Rehabilitation 80(8):945-950.
3. Ludewig, P. et al. (2009) Journal of Orthopedic Sport Physical Therapy, 39(2):90-104.
4. Lin, J. et al. (2010) Spine, 35(11): 1151-1157.
5. Wu, G. et al. (2005) Journal of Biomechanics, 38(5):981-992.
6. Karduna, A. et al. (2001) Journal of Biomechanical Engineering, 123(2):184-190.

DISCLOSURE STATEMENT

The authors have no conflicts of interest to disclose.

Comparing Measurements of Shoulder Motion by a Physician, by an Occupational Therapist with a Goniomenter and by Motion Capture

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INTRODUCTION:

Brachial plexus birth palsy (BPBP) occurs in 0.4-4.6 out of 1000 live birth and most commonly affect the C5 and C6 nerve roots [1]. Monitoring passive shoulder motion over time is important in children with BPBP as loss of passive external rotation (ER) is a sign of glenohumeral (GH) dysplasia and is an indication for further assessment and intervention [2]. Cross-body adduction (CBA) has recently been associated with scapular winging, a frequent complaint among children with BPBP and their caretakers [3]. The purpose of this study was to test the precision and accuracy of the measurements of shoulder ER and CBA by 1) a pediatric hand surgeon's visual estimate (MD), 2) an occupational therapist using a goniometer (GM) and a motion capture system (MC - considered the gold standard).

CLINICAL SIGNIFICANCE

Shoulder ER and CBA are frequently estimated by surgeons at each office visit to monitor joint range of motion. However, the accuracy of these clinical estimates has not been determined. This study aimed to determine the accuracy of range of motion visually estimated by MD and GM measurements taken by an occupational therapist to inform clinical practice.

METHODS

Twenty-six children (9.9 ± 3.2 years) with BPBP were recruited for this study. A pediatric hand surgeon (MD) visually estimated each child's passive humerothoracic (HT) ER and passive GH CBA during the patient's clinical visit. Then an occupational therapist measured the same parameters using a goniometer (GM) while motion capture data (MC) was simultaneously collected. A motion capture system (Vicon, Oxford, UK) was used to measure the 3D orientations of the trunk, scapula, and humerus segments as subjects held their arms in static positions. Joint angles at each position and displacements from neutral were calculated using a modified globe method. To look at agreement across measurements for the motion of HT ER and GH CBA, six Bland-Altman plots were created - MD vs. MC vs. GM. A one-way ANOVA was completed to test for statistical differences between groups.

RESULTS

Comparison of clinical estimates, goniometer measurements and motion capture measurements demonstrated no significant differences ($p = 0.751$) for GH CBA, while all HT ER measures were significantly different from each other (Table 1).

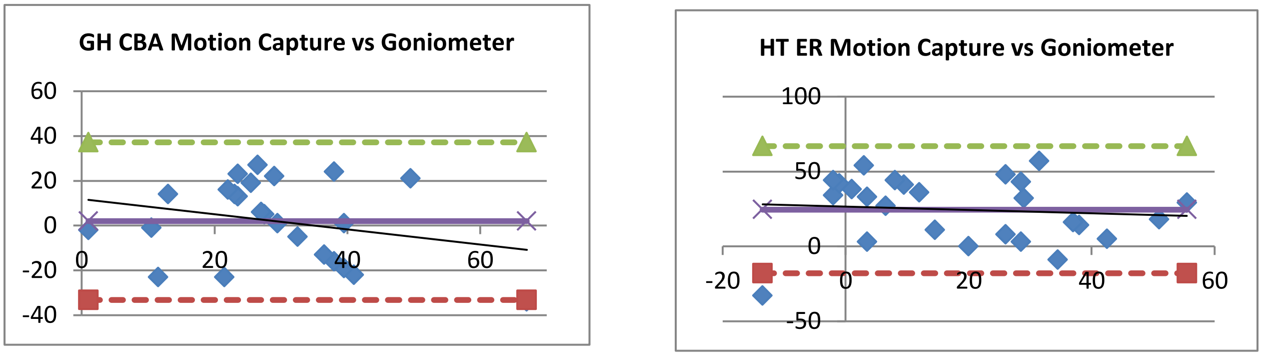
The GH CBA comparisons exhibited good accuracy with the Bland-Altman plots with the mean difference near zero (Figure 1). However, the same comparisons demonstrated poor precision with a minimum of 20° degrees deviations from neutral (Figure 1). For HT ER, there was poor precision and poor accuracy. Looking at precision using MC as the gold standard, both GM and the MD measurements overestimated HT ER by 9 and 24 degrees respectively (mean difference). There was also poor accuracy, with the smallest 95% CI across all comparisons being ± 24 degrees.

Table 1: Average GH CBA and HT ER by measurement type

Joint Angle	Motion Capture	Goniometer	Clinical Estimate
GH Cross-body Adduction	27.9 ± 18.4	27.3 ± 20.4	30.0 ± 14.4
HT External Rotation	6.7 ± 22.0*	15.6 ± 23.2*	31.2 ± 20.2*

***p<0.05 for all HT ER comparisons**

Figure 1: Selected Bland-Altman graphs comparing motion capture to occupational goniometer measure for HT ER and GH CBA.



DISCUSSION

The lack of accuracy and precision for HT ER is not surprising and suggests measurements taken through different methods should not be used interchangeably. When performing shoulder external rotation, BPBP patients often rotate and laterally flexed their trunk and extend their elbows. These compensations are difficult for a clinician to control and ultimately reduce the measurement's accuracy. Furthermore, to perform this measurement, a clinician has to have a true overhead view of the patient's trunk to accurately align the humerus with the thorax. A similar situation exists when measuring GH CBA. For this motion, the clinician has to simultaneously stabilize the scapula and position the humerus. They then have to release the scapula and then correctly position the goniometer while ensuring the patient hasn't moved the trunk or humerus. Increasing the difficulty of this measurement is that the scapula rarely lies in a cardinal plane. Only motion capture allows for visualization of the patients' multi-segmental compensations and provides an accurate representation of HT ER and GH CBA.

Visual estimates and goniometer measures of external rotation, performed according to current standard of care, were not consistent in this cohort. Clinicians should ensure that the same individual and same measurement technique are used to assess joint range of motion over time, as well as before and after interventions. Lastly, motion capture is an important tool that could improve accuracy and precision of shoulder motion measurements and should be considered for routine clinical care for patients with BPBP.

REFERENCES

1. Hoeksma A. et al. (2000) Clin Rehabil. 14(5):523-6.
2. Waters PM. et al. (1998) J Bone Joint Surg Am. 80(5):668-77.
3. Russo SA et al (2015) J Pediatr Orthop. 35(3):240-5.

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EVALUATION OF SHOULDER STRETCHING WITH AND WITHOUT SCAPULAR STABILIZATION IN CHILDREN WITH BRACHIAL PLEXUS INJURIES

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INTRODUCTION

One out of every 1000 live births results in brachial plexus birth palsy (BPBP) that causes sustained impairments, including reduced strength and range of motion, scapular winging and glenohumeral (GH) dysplasia. A clinical indication of GH dysplasia is loss of passive external rotation. Early management of BPBP is focused on maintaining full passive range of motion of all involved joints. Typically, a home stretching program is initiated around one week of age [1,2] and supplemented with formal therapy [1,2]. The mainstay of treatment has been passive GH external rotation stretching with scapular stabilization [2,3], but passive GH stretching in other planes, such as abduction, while maintaining scapular stabilization has also been recommended [3]. However, the effectiveness of scapular stabilization during shoulder stretching has not been measured. We hypothesized that scapulothoracic (ST) displacement would be decreased and GH displacement would be increased for both external rotation and abduction shoulder stretches with scapular stabilization compared to the same stretches without scapular stabilization.

CLINICAL SIGNIFICANCE

This study assessed the efficacy of scapular stabilization during shoulder joint stretching for children with BPBP. This information will provide objective evidence to inform treatment recommendations for the pediatric BPBP population.

METHODS

Twenty-six children (9.9 ± 3.2 years) with BPBP were recruited for this study. A motion capture system (Vicon, Centennial, CO) collected coordinate data in each of the following static positions: neutral with the arm resting by the side, external rotation with scapular stabilization, external rotation without scapular stabilization, abduction with scapular stabilization, and abduction without scapular stabilization. The external rotation and abduction stretches with and without stabilization were performed both by an occupational therapist with more than 20 years of experience in pediatric occupational therapy and by the child's caretaker. Manual pressure was applied to the scapula to stabilize it against the rib cage and block upward rotation and scapular winging (Figure 1). The therapist provided instruction to the caretaker, and

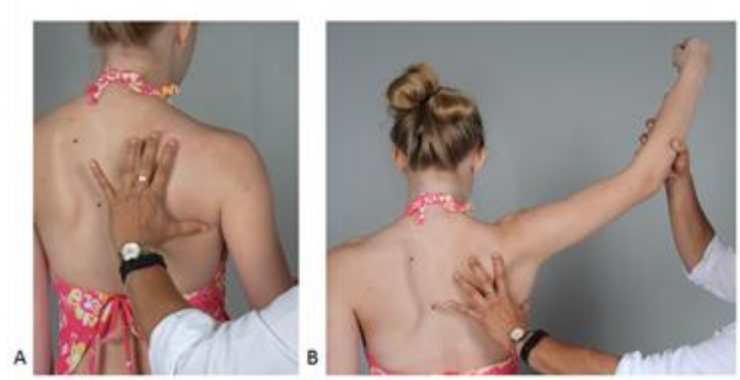

Figure 1: Manual scapular stabilization for (A) external rotation and (B) abduction stretching.

the caretakers' were allowed as many practice stretches as they needed to feel comfortable performing the maneuver.

Scapulothoracic joint angles were calculated using a helical method [4], and GH joint angles were calculated using a modified globe method [5]. Joint angular displacements between each tested position and the neutral position were calculated. One-way analyses of variance (ANOVA) with repeated measures were run for each position. The factor levels consisted of no scapular stabilization, manual scapular stabilization by an occupational therapist, and manual scapular stabilization by a caretaker. The dependent variables were ST and GH external rotation displacement for the external rotation position and ST upward rotation and GH abduction for the abduction position. In the event of a significant Wilk's Lambda, Bonferroni post hoc comparisons were made to determine which stabilization conditions were significantly different. A Bonferroni multiple comparisons correction was applied to the ANOVAs to account for testing two positions, external rotation and abduction ($\alpha = 0.025$).

RESULTS

There were no significant differences in either the ST or GH joint external rotation displacement; however, both ST upward rotation and GH abduction were significantly decreased with scapular stabilization performed by the therapist and the caretaker (Table 1).

Table 1: Joint angular displacements ± standard deviation in degrees along with the significance levels of the analyses of variance and Bonferroni post hoc comparisons.

Position	Joint Displacement	No Stabilization	OT Stabilization	Caretaker Stabilization	Wilk's Lambda	Post Hoc Comparison
External Rotation	ST ER Disp	4.5 ± 6.9	5.6 ± 7.0	7.9 ± 7.9	0.076	-
	GH ER Disp	35.3 ± 16.6	30.1 ± 12.6	31.0 ± 23.1	0.071	-
Abduction	ST Up Rot Disp	58.4 ± 14.1†‡	47.0 ± 13.9†	45.6 ± 16.1‡	<0.000*	†<0.000, ‡<0.000
	GH Elev Disp	42.7 ± 18.6†‡	30.0 ± 18.2†	29.3 ± 21.9‡	<0.000*	†<0.000, ‡0.001

DISCUSSION

The findings of this study indicate that scapular stabilization may be detrimental to passive stretching of the GH joint in children. During abduction stretching with scapular stabilization, ST upward rotation was successfully restricted, but the GH joint abduction stretch was less than when the abduction stretch was performed without scapular stabilization. For external rotation stretching, there were no differences in ST or GH joint stretches. The results were similar for scapular stabilization performed by the occupational therapist and by the caretakers. Based on the findings of this study, passive stretches without scapular stabilization are recommended for children with BPBP.

REFERENCES

1. Soldado, F. et al. (2005) Journal of Pediatric Orthopaedics, 25(5): 666-70.
2. Bhardwaj, P. (2013) Journal of Hand Surgery, 38(8): 1557-66.
3. Duff, S. et al. (2007) Clinical Biomechanics, 22(6): 630-8.
4. Woltring, H. et al. (1985) Journal of Biomechanics, 18(5): 379-89.
5. Pearl, M. et al. (1992) Journal of Shoulder and Elbow Surgery, 1(2):113-118.

DISCLOSURE STATEMENT

None of the authors have conflicts of interest to disclose.

HEAD KINEMATICS AND HEAD-TRUNK COORDINATION IN PEOPLE WITH AND WITHOUT VESTIBULAR HYPOFUNCTION

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INTRODUCTION

People with vestibular hypofunction acutely restrict head motion to try to reduce dizziness and nausea. This may result in an alteration of normal decoupling of head motion from body motion while walking [1, 2]. Compared to individuals with normal vestibular function, people with vestibular hypofunction also reorganize their head movements differently and sometimes less efficiently when performing gait and balance related activities [2]. The magnitude and persistence of these deficits is unclear. This study aimed to quantify head kinematics and head-trunk coordination in people recovering from unilateral vestibular deafferentation (resection of an acoustic neuroma) compared to healthy individuals without such deficits. We hypothesized that individuals with unilateral vestibular hypofunction would demonstrate altered head motion and head-body coupling at 1.5 months post-surgery compared to healthy individuals.

CLINICAL SIGNIFICANCE

A key component to recovery from peripheral vestibular deficits is regular exposure to head movements that induce retinal slip and regulate recovery of gaze stability through vestibular ocular reflex adaptation and other mechanisms [3]. To target and dose interventions appropriately, quantification of the extent of the problem and the particular areas of deficit are required. The results of this study will direct patient-centric treatment strategies to optimize recovery of head kinematics and head-trunk coordination in individuals with vestibular deficits.

METHODS

Thirteen adults with unilateral vestibular hypofunction (4 males, aged 48 ± 13 years, mean 41 ± 5 days post acoustic neuroma resection) and 20 healthy individuals without vestibular hypofunction (9 males, aged 34 ± 12 years) participated. Participants wore Opal sensors (APDM Inc, Portland OR), sampling at 128 Hz, on their forehead, sternum and waist while performing four walking tasks: i) walking 10 m at their usual pace while looking side to side every three steps (Functional Gait Assessment [FGA]-3), ii) performing a rapid 180° turn when verbally cued midway through the 10 m walk (FGA-5), iii) the Timed Up and Go (TUG) and iv) the 2-minute walk (2MWT) tests. Peak head rotation amplitude, peak head rotation velocity, body rotation lag and amount of head-trunk coupling (where values close to 0 indicate head rotation independent to trunk motion and values close to 1 indicate en bloc turns) were derived from sensor data. Raw data were filtered using a 6 Hz low pass filter and processed data visually inspected prior to analysis. Body turns greater than the usual degree of trunk rotations during gait, as identified from normal walking trials, were used for analysis. All tasks except FGA-5 involved multiple turns, so the average value for each individual for each task was used in analysis. Differences in outcomes between individuals with and without vestibular hypofunction were compared using independent samples t-tests or Mann-Whitney tests for outcomes with non-normal distributions or where the assumption of homogeneity of variance was violated. The Bonferroni-adjusted critical α was $\leq$.003.

RESULTS

Individuals with unilateral vestibular hypofunction had worse dynamic stability than healthy individuals at 1.5 months post-surgery (mean FGA score 20 ± 3 versus 30 ± 0, respectively, p<.001). Those with vestibular hypofunction demonstrated reduced head turn amplitude and increased head-body coupling for FGA-3 compared to healthy individuals (Table 1). Individuals with vestibular hypofunction also had slower head turn velocities for all tasks.

Table 1: Head kinematics and head-trunk coordination in individuals with and without unilateral vestibular hypofunction (mean ± SD).

Task	Outcome	Vestibular (n=13)	Healthy (n=20)	p-value
FGA-3	Head turn amplitude (deg)	83.5 (16.0)	113.2 (24.4)	**<.001**
	Head turn velocity (deg/s)	191.1 (77.5)	358.9 (112.5)	**<.001**
	Head-body coupling (%)	16.1 (6.4)	6.8 (5.3)	**<.001***
FGA-5	Head turn amplitude (deg)	154.8 (12.7)	154.9 (19.1)	.61*
	Head turn velocity (deg/s)	233.7 (67.2)	348.5 (98.8)	**<.001**
	Body turn lag (msec)	84.1 (58.1)	38.3 (57.3)	.03
	Head-body coupling (%)	104.6 (12.7)	109.5 (15.5)	.35
TUG	Head turn amplitude (deg)	155.2 (11.3)	166.5 (10.8)	**.003***
	Head turn velocity (deg/s)	157.2 (35.6)	242.6 (47.2)	**<.001**
	Body turn lag (msec)	59.2 (151.0)	139.3 (130.8)	.25*
	Head-body coupling (%)	98.8 (10.0)	102.5 (6.8)	.27*
2MWT	Head turn amplitude (deg)	155.7 (11.4)	157.3 (12.4)	.71*
	Head turn velocity (deg/s)	168.0 (38.3)	220.4 (52.2)	.004
	Body turn lag (msec)	138.1 (130.3)	149.2 (113.3)	.80
	Head-body coupling (%)	103.4 (12.4)	106.5 (7.3)	.53*

Statistically significant between-group difference. *Determined by the Mann-Whitney test.

DISCUSSION

While the central nervous system facilitates recovery of gaze and postural stability following unilateral vestibular deafferentation, the extent and time course to which individuals normalize their head kinematics and head trunk dissociation have not previously been examined. We found that head angular amplitude and velocity deficits persisted in individuals 1.5 months post acoustic neuroma resection compared to individuals without vestibular hypofunction. Since head movement and retinal slip error are critical to recruiting oculomotor strategies to improve gaze stability [3], such restrictions may postpone and limit the extent of gaze stabilization recovery. Future research is needed to understand the time course of recovery of head and trunk kinematics following vestibular deafferentation, and how movement dosage affects recovery.

REFERENCES

1. Cavanaugh, J. et al. (2005) Journal of Rehabilitation Research & Development 42(2):191-98.
2. Mijovic, T. et al. (2014) Otology & Neurotology 35(10):e348-e357.
3. Schubert, M.C. and Zee, D.S. (2010) Restorative Neurology & Neuroscience 28(1):9-18.

DISCLOSURE STATEMENT

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Effect of Cervical Decompression Surgery on Gait in Adult Cervical Spondylotic Myelopathy Patients

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INTRODUCTION

Cervical spondylotic myelopathy (CSM) is a neurologic condition resulting from spinal cord compression due to degenerative changes within the cervical spine.[1] This can include disc herniations, disc osteophyte complexes, ligamentum flavum hypertrophy, spondylolisthesis, or cervical kyphosis.[2] CSM is most common after the age of 50 years, but the age of onset is variable depending on the degree of congenital spinal canal narrowing and other factors.[1] Gait imbalance is a frequent symptom of CSM, and has been reported to be improved by surgical intervention. Clinical studies have determined that individuals with CSM have a slower gait speed, prolonged double support duration and reduced cadence compared to healthy controls.[3] Previous studies also identified reduced knee flexion during the swing phase in the early stages of the disease, and in more severe cases, decreased ankle plantar flexion at the terminal stance and reduced knee flexion during loading response.[4] While there may be some debate as to when patients with radiographic cervical stenosis should undergo decompressive surgery, most surgeons will agree on surgery for patients with moderate or severe myelopathy.[5] No matter what specific operation one chooses for a patient with CSM, the primary objective is to decompress the spinal cord.[5] The purpose of this study was to evaluate the effect of cervical decompression surgery on the biomechanics of the lower extremities and spine during gait in patients with CSM before and after surgical intervention and to compare these parameters to an asymptomatic group.

CLINICAL SIGNIFICANCE

Cervical decompression surgery may improve gait performance on cervical spondylotic myelopathy patients.

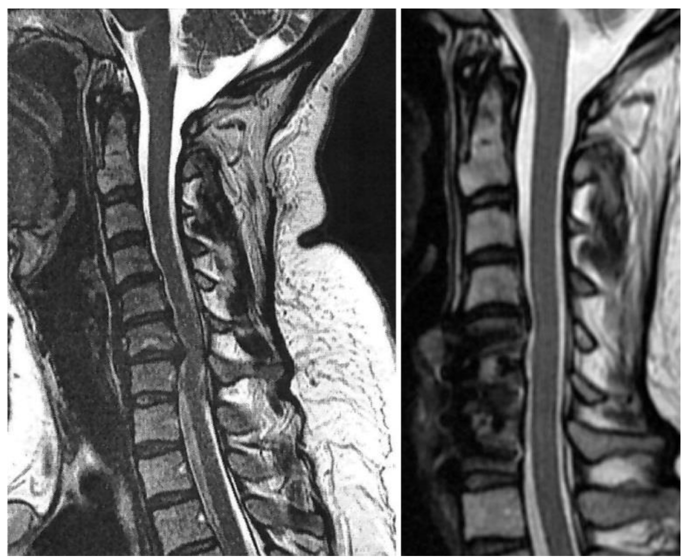

Figure 1. Preoperative and postoperative MRI of a myelopathic patient

METHODS

Eight subjects with CSM who have been deemed appropriate surgical candidates performed gait analysis a week before (Pre) and 3 months after the surgery (Post3). Twenty healthy volunteers served as a control group. Fifty-one reflective markers (9.5 mm diameter) were incorporated to collect full body three-dimensional kinematics using 10 cameras (VICON, Denver, CO) at a sampling rate of 100 Hz. Ground reaction forces (GRFs) were measured using three parallel force plates (AMTI, Watertown, MA). The patient walked barefoot at his/her self-selected speed along a 10 m walkway. Five trials were recorded during each session. Spine and lower extremity kinematic and vertical GRF were measured and recorded. One-way ANOVA with Bonferroni Post Hoc analyses was used to determine differences in gait patterns in adult CSM patients before and after their surgery and compared to the healthy group.

RESULTS

After cervical decompression surgery, CSM patients had significantly faster walking speed (Pre: 0.82 vs Post3: 1.03 m/s, p=0.050), longer step (Pre: 0.48 vs Post3: 0.60 m, p=0.013) and stride length (Pre: 0.98 vs Post3: 1.14 m, p=0.050). A significantly smaller ankle plantar flexion ROM (Pre: 29.46 vs Post3: 20.87 deg, p=0.033) was seen during the stance phase. The first peak of the vertical GRF was also found to occur earlier (Pre: 21.05 vs Post: 17.90 % of gait cycle, p=0.050) after cervical decompression surgery (Figure 2). In comparison to the control group, CSM patients preoperatively presented with a significantly slower gait speed (0.24 m/sec; p=0.037), decreased step length (0.11 m; p=0.014), stride length (0.20 m; p=0.019) and increased step width (0.05 m; p=0.001). Moreover, CSM patients presented with a longer double support time (0.01 s; p=0.050). Cadence, stride and stance times, swing-Stance ratio, and single support time were not found to be statistically different (Table 1). Furthermore, CSM patients showed a significantly larger ankle (5º; p=0.024) ROM and smaller knee (15º; p=0.050) ROM in the sagittal plane, along with greater ankle (2º; p=0.050) ROM in the coronal plane. Moreover, preoperative CSM patients presented with a lower vertical GRF 2nd peak (0.09% of BW; p=0.050), later 1st peak (3% of gait cycle; p=0.048) and valley (3% of gait cycle; p=0.049) in comparison to the healthy group (Figure 2). Minor differences in gait found between the post-surgical CSM patients in comparison to the control group. CSM patients showed a significantly larger hip (4º; p=0.038) and smaller pelvis (5º; p=0.015) ROM in the coronal plane.

DISCUSSION

Cervical decompression surgery improved the gait pattern in patients with CSM. Based on our preliminary results, surgical decompression resulted in faster walking speeds with longer steps with increase in spine and lower extremity function and efficiency. Cervical spondylotic myelopathy patients walk slower with reduced trunk and lower extremity function and efficiency in comparison to an asymptomatic group. Post-operative CSM patients actually had similar walking patterns in comparison to an asymptomatic group. Formal gait and motion analysis can provide an objective method to assess the impact of spinal cord compression on a patient's gait and lower extremity function and also monitor the subsequent improvement postoperatively.

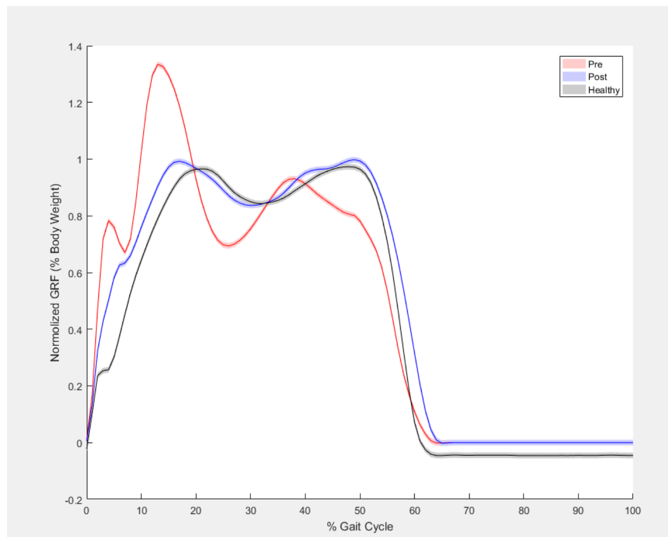

Figure 2. Vertical ground reaction forces during gait from a representative patient with cervical myelopathy pre- and post-surgery and healthy adults.

REFERENCES

1. Rhee JM, et al. *Spine J* **34**(9), 890-5, 2009.
2. Emery SE, et al. *J Am Acad Orthop Surg* **9**(6), 376-88, 2001.
3. Malone A, et al. *Eur Spine J* **21**(12), 2456–66, 2012.
4. Tetreault L, et al. *Neurosurgery* **77**(4),S51-67, 2015.
5. Geck MJ, et al. *Orthop Clin North Am J* **33**(2), 329-48, 2002.

DISCLOSURE STATEMENT

R. Haddas, K. Ju, R. Arakal and T. Belanger have no conflicts of interest to disclose.

A COMPREHENSIVE METHOD TO MEASURE THREE-DIMENSIONAL BREAST MOTION DURING PHYSICAL ACTIVITY

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INTRODUCTION

Breast motion research aims to understand how the breasts move in order to aid development of apparel that appropriately minimizes breast motion. The majority of the studies in this field quantify breast motion as motion of the nipple relative to the sternal notch [1,2,3]. A more recent study examining breast motion in sports bras found that five markers placed at set distances from the nipple demonstrated substantially different displacements compared to those of the nipple marker [4]. While these findings illustrate that breast motion is more complex than nipple motion, they measure only a small region of the bra, and are consequently of little benefit to bra design. The purpose of this study was to develop and assess an objective method to comprehensively measure 3-D bra/breast motion with the end goal of informing bra design.

CLINICAL SIGNIFICANCE

The results from this study may aid the development of clothing that effectively controls breast motion while maximizing comfort during physical activity, and this may ultimately increase breast health and women's participation in physical activity.

METHODS

Six females (age: 26 ± 7 years) were recruited with institutional IRB approval. All participants were healthy, capable of performing physical activity, and appropriately fitted with a Reebok International Ltd. minimally supportive, seamless bra. Three participants were fitted with a small seamless bra and three participants were fitted with a large seamless bra. A total of 54 retro-reflective markers were used to track breast motion. Fifty-one markers were placed across the anterior straps and body of the bra in a grid pattern, which was used to standardize marker placement across different bra sizes (Fig. 1). In addition, three markers were used to define the torso [2]. A 13-camera motion analysis system (Motion Analysis Corp., Santa Rosa, CA, USA) captured the motion of the 54 markers as each participant ran on a treadmill. All marker trajectories were tracked in Cortex (Motion Analysis Corp., Santa Rosa, CA, USA) and processed using custom-written software. The locations of minimum and maximum marker displacements, marker velocities, link stretch

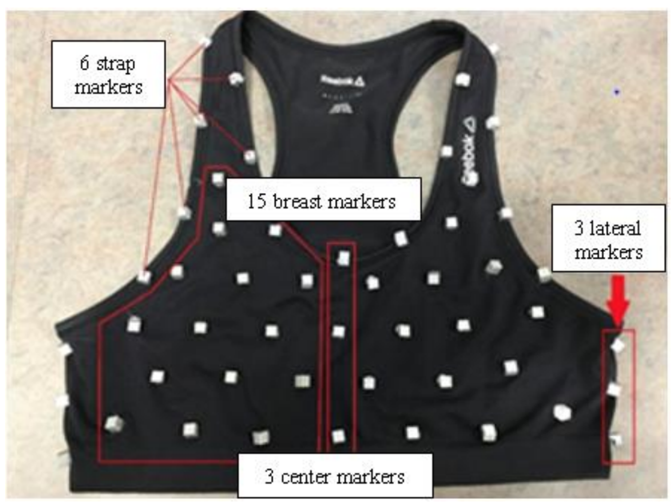

Figure 1: Grid pattern of 51 markers.

(“link” refers to the space between adjacent markers), and link stretch velocities were identified to quantify breast motion.

RESULTS

Maximum marker displacements and marker velocities on both bra sizes were located in the lower breast region. Average maximum link stretch and link stretch velocity for both bras were located at the bra/shoulder-strap interface. Average minimum values for all measures were roughly the same independent of bra size (Table 1). However, average maximum values for the large bra were roughly double that of average maximum values for the small bra across all measurements (Table 1). For both bra sizes, maximum marker motion was also substantially greater than that of the marker closest to the nipple.

Table 1: The minimum and maximum (mean ± SD) marker displacement, marker velocity, link stretch, and link stretch velocity values during running.

Measurement	Bra Size	Minimum	Maximum	Marker Closest to Nipple
Marker Displacement (mm)	**Small**	13.8 ± 7.2	27.1 ± 7.5	22.1 ± 9.0
	Large	15.3 ± 4.6	42.4 ± 11.9	39.3 ± 10.8
Marker Velocity (mm/s)	**Small**	92.7 ± 24.3	333.4 ± 106.7	221.2 ± 111.5
	Large	102.7 ± 27.9	557.3 ± 210.9	438.9 ± 112.0
Link Stretch (mm)	**Small**	0.8 ± 0.2	9.4 ± 2.6	n/a
	Large	0.89 ± 0.0	15.9 ± 1.8	n/a
Link Stretch Velocity (mm/s)	**Small**	7.1 ± 2.4	115.3 ± 40.6	n/a
	Large	7.5 ± 0.8	185.3 ± 19.1	n/a

DISCUSSION

This study found that maximum breast displacement and velocity occurred in the lower breast region, a region not formerly analyzed with previous marker sets [4]. Furthermore, this study found that link stretch and link stretch velocity were greatest in the strap region of the bra. These measurements may be important for designing innovative clothing that minimizes breast motion during physical activity. Finally, the results from this study support previous findings that marker motion at the nipple differs substantially from marker motion located at other regions on the breast.

REFERENCES

1. Bridgman, C. et.al. (2010) Journal of Applied Biomechanics, 26(4):465-472.
2. Haake, S. et.al. (2011) Sports Engineering, 14(1):49-56.
3. Scurr, J. et.al. (2009) Journal of Applied Biomechanics, 25:322-329.
4. Zhou, J. et.al. (2012) Textile Research Journal, 83(14):1500-1513.

ACKNOWLEDGMENTS

The authors wish to thank Reebok International Ltd. for providing the bras used for testing.

DISCLOSURE STATEMENT

The authors have no conflicts of interest to disclose.

Effect of Load Carriage on Lower Extremity Coordination Variability During Constant Speed Treadmill Marching in ROTC Cadets

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INTRODUCTION

Load carriage has been, and continues to be, a primary occupational task that characterizes military service as evidenced by its inclusion in several field fitness tests for new and continuing service members. The postural effects of load carriage have long been documented in a variety of populations, including military personnel [1]. The high volume (e.g. mileage) and intensity (e.g. heavy loads) that is often associated with military style marches has been associated with considerable changes in trunk posture and gait kinetics, a possible source of the high rates of cumulative trauma musculoskeletal disorders that is observed in this population, especially among new recruits [2, 3]. While preventative measures have been developed through technologies for better carriage systems in an attempt to address many of these injuries, these technologies remain in conflict with tactical requirements that are present for active duty personnel. Research has identified changes in stride frequency as load increases and in stride length and frequency during forced march pacing; however, to date there is no research that examines the military standard pacing necessary to pass the 20 mile march on lower extremity coordination variability. Therefore, the primary purpose of the present study was to examine the effect of constant speed on lower extremity coordination variability under various loaded conditions in ROTC Cadets.

CLINICAL SIGNIFICANCE

Understanding potential load carriage effects on lower extremity coordination variability during military style marching may assist in creation of strategies to reduce the development of CT MSK disorders in this population.

METHODS

Ten (10) healthy, male and female Air Force Senior Reserve Officer Training Corps (SROTC) cadets (mean age=19.8 ± 2.2 years) were recruited for this study. All subjects were free of pathologies known to influence load carrying and gait patterns for at least the past year. All participants signed informed consent forms according to Louisiana Tech University's policies and procedures. Participants were dressed in identical spandex clothing and wore commercial running shoes. Prior to all sessions, participants had retro-reflective markers placed on bony landmarks to calculate joint angles. Two-dimensional sagittal plane kinematics were collected through a single Basler Scout camera recording at 100 Hz and processed through MaxTRAQ2D (Innovision Systems, Inc., Columbiaville, MI, USA). All trials were set to a constant speed of 1.79 $m*s^{-1}$ (4 miles per hour) to simulate the minimal speed that is required of military personnel to pass the 20 mile march (time limit: 5 hours) physical fitness test, which includes a military pack with a 22.7 kg (50 lbs) load. Participants were blinded to the treadmill speed during all trials. Four randomized trials were completed under the following conditions: no pack, pack with no load, 11.5kg, and 23 kg for each trial. All trials utilizing a pack were completed while carrying a military issue All-purpose Lightweight Carrying Equipment (ALICE). Each trial lasted 3 minutes in length, with data collected during the final 60 seconds of each trial. Participants were also given 3 minutes of rest between each load condition. A custom Matlab program (The Mathworks, Inc., Natick, MA, USA) was utilized to determine the continuous relative phase (CRP) ratios and deviation phase (DP) of the thigh-shank and shank –foot during each condition, with sagittal plane joint angles and velocities calculated the final 60 seconds of each 3 minute trial. Separate 1x4 Repeated Measures ANOVA were conducted with follow-up Pairwise Comparisons.

RESULTS

Results indicated that there were no significant differences found in the CRP $_{mean}$ or DP of the thigh-shank between any of the conditions; however, results indicated a significant load effect on the DP for the shank-foot ($\Lambda_{Wilks'}$ = 0.195, $F(3,7) = 9.625$, $p = 0.007$, η^2= 0.805). Follow-up analyses indicated that participant displayed a significantly lower shank-foot DP during the 23kg load condition as compared to all other conditions (Figure 1)

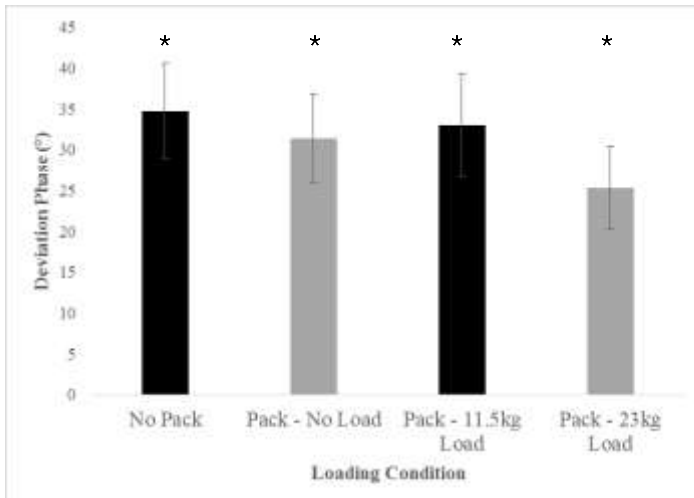

Figure 1: Shank-Foot Deviation Phase during Constant Speed Marching.
* Pack – 23kg Load: Significantly ($p \leq 0.013$) lower than all other conditions.

DISCUSSION

The present study indicates that while load magnitude causes minimal changes in thigh-shank deviation phase during acute bouts of exercise, there is a significant loss in shank-foot deviation phase at those loads used by the U.S. Army during constant speed marches for new recruits. Active duty military personnel, especially new recruits, routinely suffer from very high rates of cumulative trauma musculoskeletal disorders such as medial tibial stress syndrome [5]. Among other factors, the reduction in lower extremity coordination variability is an area that warrants further investigation as a potential injury prevention pathway for individuals undergoing load carriage as part of their occupation or leisure.

REFERENCES

1. Attwells, R. L. et al. (2006). Ergonomics, 49(14), 1527-1537.
2. Birrell, S. A. et al. (2010). *Applied Ergonomics*, *41*(4), 585-590.
3. Chow, D. H. K. et al. (2011). *International Journal of Industrial Ergonomics*, *41*(5), 530-535.
4. Beekley, M. D. et al. (2007). *Military Medicine*, *172*(6), 592-595.
5. Sharma, J. et al. (2011). *Gait & Posture,* 33(3), 361-365.

Daily Schedule: Friday 26 May 2017

GCMAS 2017 Annual Meeting

University of Utah Guest House

7:00A – 12:00P

TIME	TOPIC & ACTIVITY	ROOM
7:00A – 10:00A	Registration	Lobby
7:00A – 7:45A	Breakfast Session: Commission for Motion Analysis Accreditation (CMLA), Gordon Alderink, PT, PhD	Douglas Ball Room
8:00A – 9:30A	Podium Session 7 – Cerebral Palsy III	Douglas Ball Room
9:30A – 10:00A	Break	
10:00A – 11:30A	Podium Session 8 – Data Methods/Modeling	Douglas Ball Room
11:30A – 12:00P	Awards Presentations	Douglas Ball Room

Breakfast Session

GCMAS Council Meetings

Friday, May 26th
7:00A - 7:45A

Council Chairs:

Education Jenna Yentes	Solicitation and selection of instructional courses to be offered at the annual Society meeting, encouraging student participation, and other educational endeavors undertaken by the Society
Research Jinsup Young	Promoting, advancing, and supporting research in the field of movement analysis
Standards Howard Hillstrom	Development of standardized approaches for clinical movement analysis including all aspects of data acquisition, reduction, utilization, interpretation, presentation, and communication
Reimbursement Sylvia Öunpuu	Providing resources to assist members in obtaining payment for motion analysis services
Membership Kirsten Tulchin-Francis	Implementing membership initiatives and responsibly representing the needs of the membership; maintenance of membership records, membership initiatives, and other matters relating to membership participation in Society activities
Awards Aviva Wolf	Development and implementation of a Society awards system designed to recognize outstanding individual and group achievements related to gait and human movement analysis in clinical and research settings
Communications Braden Romer	Developing and maintaining the society's web page and list server, publishing the newsletter and other communication endeavors undertaken by the society

If you are a GCMAS member and would like to be involved in the society, please speak with one of the council chairs, we are always looking for new members to help shape the future of our organization.

Podium Session #7

CEREBRAL PALSY III

MODERATED BY: **Susan Rethlefson, PT:** Motion Analysis Laboratory
Children's Hospital Los Angeles, Los Angeles, CA

Kristan Pierz, MD: Medical Director, Center for Motion Analysis
Connecticut Children's Medical Center, Glastonbury, CT

1. **Hip Power and "Stiff Knee" Gait: A Tool for Identifying Appropriate Candidates for Rectus Transfer**
   Tasos Karakostas, Brunno Moreira, Vineeta Swaroop, Luciano Dias

2. **Does Rectus Femoris Transfer Mitigate Iatrogenic Kinematics After Distal Femoral Extension Osteotomy and Patella Tendon Advancement in Individuals with Cerebral Palsy**
   Elizabeth Boyer, Tom Novacheck, Michael Schwartz

3. **Effects of Multilevel Surgery with Hamstring Lengthening in Ambulatory Children with Cerebral Palsy**
   Brian Chen, John Henley, Julieanne Sees, Mutlu Cobanoglu, Kenneth Rogers, Chris Church, Nancy Lennon, Freeman Miller

4. **Elevated Forces of Spastic Semitendinosus Muscle Due to Inter-Muscular Interactions in Children with Cerebral Palsy**
   Filiz Ates, Yener Temelli, Can A. Yucesoy

5. **Best Leg Length Discrepancy in Children with Spastic Hemiplegic Cerebral Palsy**
   Jason Rhodes, Allison Frickman, Steven Gibbons

6. **Effect of Selective Dorsal Rhizotomy On Ankle Joint Stiffness**
   Filiz Ates, Kenton K. Kaufman

7. **Comprehensive Long-term Outcomes 10-17 Years After Selective Dorsal Rhizotomy**
   Meghan Munger, Nanette Aldahondo, Linda Krach, Tom Novacheck, Michael Schwartz

8. **How Does Hemispherectomy Affect Gait?**
   Alexis Gerk, Amy Winter Bodkin, Richard Pimentel, James Carollo, Joanna Roybal, Michael Handler, Brent O'Neill, Kevin Chapman, Ilana Neuberger, Michael Dichiaro, Zhaoxing Pan, Frank Chang

HIP POWER AND "STIFF KNEE" GAIT: A TOOL FOR IDENTIFYING APPROPRIATE CANDIDATES FOR RECTUS TRANSFER

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INTRODUCTION

Stiff knee gait is a common gait abnormality in ambulatory children with cerebral palsy. It presents with diminished knee flexion during the swing phase of gait, and delayed peak knee flexion. Rectus femoris over-activity is considered the primary cause of the deviation and rectus femoris transfer surgery (RFTS) is performed to decrease the muscle's ability to extend the knee [1]. Knee flexion improvement following RFTS is believed to be due to reduction of the muscle's knee extension moment while preserving its hip flexion moment [2].

However, RFTS outcomes are inconsistent, in part, due to lack of identifying and understanding the effective predictors for positive outcomes [3]. Knee flexion velocity at toe-off and peak knee flexion in swing have been found to be correlated [4]. Furthermore, normal gait simulation studies have shown that decreased hip flexion moment results in reduced peak knee flexion in swing [5]. While the triceps surae power generation is the major contributor for propulsion during normal walking, in stiff knee gait loss of distal control is observed and, therefore, power generation must be reverse-generated from the hip [6]. Consequently, although to the best of our knowledge not yet investigated, peak hip power generation in late stance or early swing may be one of the most important characteristic features of the stiff limb associated with improvements in knee flexion during swing.

Therefore, with the use of instrumented 3D gait analysis, the purpose of this study was to investigate the potential of the hip power characteristics in predicting the outcome of RFTS in children with cerebral palsy diagnosed with stiff-knee gait.

CLINICAL SIGNIFICANCE

If certain hip power characteristics are determined to be good predictors of RFTS outcome, the effectiveness of selecting appropriate candidates for the procedure will be improved.

METHODS

Sixteen children (mean age 12.4yo, range 8-20.2yo; 12 males, 4 females) with spastic cerebral palsy, GMFCS I /II who underwent RFTS (16 unilaterally, 4 bilaterally, i.e., 20 limbs total) as a swing phase surgery for stiff knee gait were evaluated retrospectively. All subjects had pre- and post-operative 3D gait analysis with successful capturing of kinetic data to compute lower extremity joint kinetics. All transfers were made to the iliotibial band by the same surgeon and all patients completed the same post-operative rehabilitation protocol successfully. Children who had traumatic brain injury were not included in the study.

Patients were classified as "Good" or "Poor" based on the RFTS outcome which, for the purposes of this study, involved a post-operative normal total knee range of motion ±2SD during gait (>47.1 degrees, or a post-operative increase in peak knee flexion in swing of more than 1SD (6.4 degrees) [7]. Then the classifications were further divided into two groups. Group I included patients whose pre-operative assessment was based on peak hip power (PHP) with a cut-off magnitude of 0.60w/kg or greater (mean PHP –2SD based on our laboratory normative database); Group II included patients whose pre-operative evaluation was based on the PHP and the time it occurred during the gait cycle (tPHP) with a cut-off time of less than 68% of the gait cycle (GC), i.e., mean tPHP+2SD based on our laboratory normative database.

RESULTS

Table 1 demonstrates the results of this study. There were 12 limbs with good outcome and 8 with poor outcome following RFTS. When the PHP and the tPHP were considered together as a predictor of the RFTS outcome, the specificity was 100%.

Table 1: Sensitivity and specificity of the Peak Hip Power (PHP) and time of Peak Hip Power (tPHP) as predictors of rectus transfer surgery outcome to treat stiff-knee gait

		SENSITIVITY	**SPECIFICITY**
	PHP	**GOOD**	**POOR**
GROUP I	PHP ≥ 0.6	11	2
	PHP < 0.6	1	6
	PERCENT	91%	75%
GROUP II	PHP ≥ 0.6 tPHP ≤ 68%GC	11	0
	PHP < 0.6 tPHP > 68%GC	1	8
	PERCENT	91%	100%

DISCUSSION

To the best of our knowledge, this is the first study investigating the PHP and tPHP as predictors of RFTS. Our results suggest that when the two parameters are factored together in the decision-making process there is the potential of identifying very successfully at the very minimum candidates who will not benefit from the procedure. Furthermore, consideration of the PHP and tPHP may allow the inclusion of RFTS as part of the single event multilevel surgery as opposed to a swing phase surgery which requires an additional operative visit.

REFERENCES

1. Perry, J. (1987) Developmental Medicine Child Neurology, 29(2):153-158.
2. Riewald, S. and Delp, S. (1997) Developmental Medicine Child Neurology, 39(2):99-105.
3. Muthusamy, K. et al. (2008) Journal of Pediatric Orthopaedics, 28(6):674-678.
4. Goldberg, S. et al. (2006) Journal of Biomechanics, 39(4):689-698.
5. Piazza, S. and Delp, S. (1996) Journal of Biomechanics, 29(6):723-733.
6. Gage, J. (1990) Clinical Orthopaedics and Related Research, 253:45-54.
7. Rethlefsen, S. et al. (2009) Journal of Pediatric Orthopaedics, 18(2):58-62.

ACKNOWLEDGMENTS

Davis, R.

DISCLOSURE STATEMENT

None of the authors have any conflict of interest to disclose.

DOES RECTUS FEMORIS TRANSFER MITIGATE IATROGENIC KINEMATICS AFTER DISTAL FEMORAL EXTENSION OSTEOTOMY AND PATELLA TENDON ADVANCEMENT IN INDIVIDUALS WITH CEREBRAL PALSY?

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INTRODUCTION

Distal femoral extension osteotomy with patella tendon advancement (DFEO+PTA) is an effective surgery for treating crouch gait [1]. However, select kinematics appear to deteriorate after DFEO+PTA, including peak knee flexion in swing [1,2] and pelvic tilt [1,3]. A rectus femoris transfer (RFT) is often implicated as a possible treatment to increase knee flexion in swing in the short- and long-term [4]. However, a RFT performed prophylactically with crouch gait treatment significantly *decreases* peak knee flexion in swing [4]. DFEO+PTA was not one of the crouch gait treatments used by Dreher et al. [4], so it is unknown if RFT may be beneficial when used in combination with DFEO+PTA. de Morais Filho et al. [2] observed decreased peak knee flexion after DFEO+PTA, suggesting that DFEO+PTA may be the culprit for loss of knee flexion, but surgeons did not also perform a RFT.

In regards to pelvic tilt, one clinical impression is to perform a RFT to prevent worsening anterior pelvic tilt after a DFEO+PTA since RF is a biarticular muscle. In such instances, RFT would serve two purposes: maintain or improve peak knee flexion and pelvic tilt. Our purpose was to determine if there are long-term beneficial effects of RFT on peak knee flexion and pelvic tilt in individuals who had a DFEO+PTA.

CLINICAL SIGNIFICANCE

Our data suggest that the RFT does not provide any long-term beneficial effects on peak knee flexion or pelvic tilt. Therefore, the merit of performing an RFT concomitantly with a DFEO+PTA for treatment of crouch should be scrutinized.

METHODS:

The IRB approved the study and written informed consent was obtained from all participants. We recruited 28 individuals (40 limbs) who had undergone a DFEO+PTA at least 8 years ago and who were presently at least 20 years old. Outcome variables included peak knee flexion in swing and mean anterior pelvic tilt. Group differences (+RFT vs. -RFT) were compared using a 2 (RFT group) × 2 (time) ANOVA. Concomitant hamstrings lengthening and psoas lengthening were compared between groups, as those surgeries may affect pelvic tilt [5,6].

RESULTS

Of the 28 individuals who had a DFEO+PTA, 10 (19 limbs) had an RFT, 3 (4 limbs) had a hamstring lengthening, and 11 (17 limbs) had a psoas lengthening. The majority of the +RFT group also had a psoas lengthening (8/10 people) and vice versa (8/11 people). One of the individuals who had a hamstring lengthening also had a RFT and psoas lengthening, whereas the other two individuals also had a RFT.

There was no significant RFT group × time interaction for peak knee flexion (p=0.89) or pelvic tilt (p=0.87), indicating that both groups responded similarly over time regardless of RFT. Nor was there a main effect for group for knee flexion (p=0.18) or pelvic tilt (p=0.28), indicating that both groups had similar knee flexion averaged across time. However, there was a main effect for time for both variables (p<0.001), indicating that peak knee flexion significantly decreased and anterior pelvic tilt significantly increased at long-term.

DISCUSSION

Our data support the findings of others that peak knee flexion decreases and pelvic tilt becomes more anterior following a DFEO+PTA [1-3]. Furthermore, concomitant RFT does not have a large effect of preventing these undesirable kinematics.

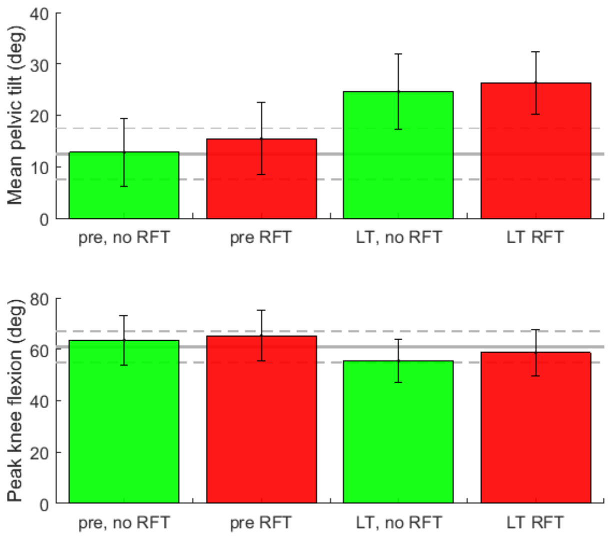

Figure. Group averages with 1SD error bars. Mean±1 SD lines are plotted in gray for typically developing.

Because of the small sample sizes and concomitant surgeries, we cannot distinguish the effects of the RFT from psoas lengthening. However, since one purpose of a psoas lengthening is to improve hip kinematics, we may conclude that neither it nor RFT are effective at preventing excessive anterior pelvic tilt in the long-term when performed in combination with a DFEO+PTA. Additionally, since long-term mean pelvic tilt of both groups falls outside two standard deviations of typical and peak knee flexion remains within one standard deviation, change in pelvic tilt may be of greater concern because of its association with low back pain [7].

Larger samples are needed to confirm or refute these findings. It is possible that RFT has a small to moderate beneficial long-term effect (more likely on knee flexion than pelvic tilt) that may be detectable with larger sample sizes.

REFERENCES

1. Stout, J. et al. (2008) Journal of Bone and Joint Surgery, 90(11):2470-84.
2. Morais Filho et al. (2016) Developmental Medicine & Child Neurology, 58:340.
3. Morais Filho et al. (2016) Developmental Medicine & Child Neurology, 58:100.
4. Dreher, T. et al. (2012) Journal of Bone and Joint Surgery, 94A(19):e142.
5. Novacheck, T. et al. (2002) Journal of Pediatric Orthopaedics, 22(2):158-64.
6. DeLuca, P. et al. (1998) Journal of Pediatric Orthopaedics, 18(6):712-8.
7. Harada, T. et al. (1993) Journal of Bone and Joint Surgery, 75(4):534-7.

DISCLOSURE STATEMENT

No author has any conflicts of interest to disclose.

EFFECTS OF MULTILEVEL SURGERY WITH HAMSTRING LENGTHENING IN AMBULATORY CHILDREN WITH CEREBRAL PALSY

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INTRODUCTION

Hamstring lengthening can effectively improve the extensibility of knee in the patient with cerebral palsy (CP) [1,2]. However, the impact of this lengthening procedure on the ambulation function was not evaluated previously with more comprehensive evaluation tools.

The aim of this study is to determine the outcomes of hamstring lengthening as a part of multilevel surgery in children with CP.

CLINICAL SIGNIFICANCE

Hamstring lengthening as a part of multilevel surgery is beneficial to ambulatory function of children with CP.

METHODS

Pre- and post-operative visits of children with spastic CP who underwent multilevel surgery with hamstring lengthening were evaluated retrospectively. Physical examination and gait kinematic data related with hamstring of pre-operative visit, short-term (12-36 months) and long-term (more than 60 months) post-operative follow-ups were analyzed.

RESULTS

This study included 194 children with spastic cerebral palsy (123 males and 71 females; Age of surgery = 11±3.8 years). In short-term evaluations (18±6.2 months; Range: 12-36 months), 314 operated limbs (from 181 children) were included. In long-term evaluations (77±15.0 months; Range: 60-114 months), 60 operated limbs (from 36 children) were included.

Knee flexor strength was weaker in 23.4%, unchanged in 60.5%, stronger in 16.1% of limbs in the short-term and was weaker in 26.8%, unchanged in 48.8%, stronger in 24.4% of limbs in the long-term. Hamstring spasticity was increased in 19.2%, unchanged in 27.8%, decreased in 53% of limbs in the short-term follow-up and increased in 45.0%, unchanged in 25.0%, decreased in 30.0% of limbs in long-term follow-up.

Knee extension passive range-of-motion had 4° short-term improvement ($p<0.05$) but returned to pre-operative level at long-term ($p=0.98$). Popliteal angle had 10° short-term improvement ($p<0.05$) and 7° long-term improvement ($p=0.01$).

Changes of Gross Motor Function Measure (GMFM) Dimension D in each pre-operative Gross Motor Function Classification System (GMFCS) level are shown in the Table 1.

Kinematics analyses of knee flexion at initial contact improved 10° in short-term ($p<0.05$) and improved 9° in long-term ($p<0.05$). Maximum knee extension during stance phase improved 9° in short-term ($p<0.05$) but returned to pre-operative position in long-term ($p=0.657$). Pelvic anterior tilt during stance phase increased 4° in short-term ($p<0.05$) but returned to pre-operative position in long-term ($p=0.282$). Hip extension during stance phase was unchanged in short-term ($p=0.36$) but increased 6° in long-term ($p=0.001$). In general, the Gait Deviation Index (GDI) improved 7.2 points in short-term ($p<0.05$) and 9.6 points in long-term ($p<0.05$). Changes of GDI in each pre-operative GMFCS level are shown in the Table 2.

Table 1: GMFM Dimension D changes in each GMFCS level. (*: statistical significance)

Follow-up	GMFCS	Cases	Pre-op GMFM (%)	Post-op GMFM (%)	p-value
Short-term (N=181)	I	28	89±11	89±7	0.963
	II	79	75±15	76±15	0.671
	III	66	43±21	45±24	0.284
	IV	8	25±26	35±23	0.271
Long-term (N=36)	I	7	92±9	92±8	0.915
	II	12	76±11	78±9	0.639
	III	17	36±20	51±23	0.022 *

Table 2: GDI changes in each GMFCS level. (*: statistical significance)

Follow-up	GMFCS	Cases	Pre-op GDI	Post-op GDI	Differences	p-value
Short-term (N=181)	I	28	73.5±12.1	81.3±14.0	7.8	0.007 *
	II	79	68.7±15.4	78.4±13.2	9.7	< 0.05 *
	III	66	57.7±13.7	64.3±16.2	6.6	< 0.05 *
	IV	8	58.8±18.3	54.1±19.8	- 4.7	0.486
Long-term (N=36)	I	7	82.4±6.4	88.8±9.0	6.4	0.178
	II	12	64.9±13.6	80.2±12.9	15.3	0.001 *
	III	17	53.9±12.4	61.9±11.8	8.0	0.022 *

DISCUSSION

Hamstring lengthening associated with multilevel surgery did not worsen strength and spasticity of hamstring in physical examination, but improved hamstring flexibility in the short-term and led to the improvements of sagittal knee, hip and pelvic kinematics that were still present in long-term follow-up analyses. Our results support what was described in previous studies [3,4]. Effects on gross motor functions based on the GMFM showed more improvements in GMFCS level III-IV and significantly improved in long-term follow-ups. However, GDI was more improved in independent ambulators especially in GMFCS level II. Further studies will be needed to clarify the direct impact from hamstring lengthening alone.

REFERENCES

1. Baumann, J. et al. (1980) International Orthopaedics, 3(4):305-309.
2. Gage, J. (1990) Current Orthopaedics Practice, 253:45-54.
3. Damiano, D. et al. (1999) Journal of Pediatric Orthopaedics, 19(3):252-358.
4. Yngve, D. et al. (2002) Journal of Pediatric Orthopaedics, 22(5):672-676.

DISCLOSURE STATEMENT

Authors have no conflicts of interest to disclose.

ELEVATED FORCES OF SPASTIC SEMITENDINOSUS MUSCLE DUE TO INTER-MUSCULAR INTERACTIONS IN CHILDREN WITH CEREBRAL PALSY

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INTRODUCTION

Spastic cerebral palsy (CP) is characterized by repetitive muscle contractions related with exaggerated stretch reflexes. The key outcome of spastic CP is compromised joint motion indicating narrow joint range of muscle force exertion and higher forces limiting joint extension. However, recent studies on spastic gracilis (GRA) [1] and semitendinosus (ST) [2] muscles showed non-zero force production from knee flexion to full extension, with the peak force encountered only in the extended positions. This unexpected lack of narrow muscle excursion was ascribed to the lack of synergistic and/or antagonistic co-activation during the experiments. However, muscle co-activation is a typical feature of daily activities and antagonistic co-activation is particularly important in CP. Direct collagenous connections exist between the epimysia of adjacent muscles, and also indirect connections are present between distant muscles via neurovascular tracts continuous with compartmental boundaries. This network causes mechanical interactions between muscles and such epimuscular myofascial force transmission (EMFT) affects human muscles *in vivo* [3]. The effects include altered strain distribution along muscle fibers, muscle force production, excursion, and shape of force-length characteristics. Therefore, EMFT caused by co-activation of other muscles can affect mechanics of spastic muscle, but has not been tested. Measuring forces of spastic ST co-activated with its antagonist vastus lateralis (VL) and also with its synergists GRA and semimembranosus (SM) muscles, our aim was to test the following hypotheses: (i) Inter-antagonistic EMFT elevates spastic ST forces (F_{ST}), (ii) reduces its excursion, and (iii) combined inter-antagonistic and synergistic EMFT further elevates those effects.

CLINICAL SIGNIFICANCE

ST muscle of patients with crouch gait is surgically lengthened to improve knee motion. However, its contribution to limited knee extension in CP is not known. Hence, revealing the mechanics of spastic ST and effects of co-activation of other muscles is essential. Intra-operatively measuring the forces directly at the muscle's tendon is a unique way to achieve this.

METHODS

The procedures were approved by a Committee on Ethics of Human Experimentation at Istanbul University, Turkey. 6 patients with spastic diplegic CP (mean (SD) = 7.7 (4.7) years at the time of hamstring lengthening surgery; GMFCS levels = II-IV) participated. Their popliteal angles (mean (SD) = 76.4° (13.1°)) indicated abnormal knee flexor tightness. Data collection succeeded general anesthesia and routine incisions to reach the distal ST tendon over which a buckle force transducer was mounted. Pairs of skin electrodes were placed over the ST, VL, GRA, and SM muscle bellies. Supramaximal activations were imposed for tetanic contractions. From 11 limbs, isometric spastic ST forces vs. knee angle (KA-F_{ST}) data were collected in 3 conditions from 120° KA to full extension: ST was activated (I) alone, (II) simultaneously with the antagonistic VL, and (III) with added synergistic SM and GRA. Changes in KA-F_{ST} and in the operational joint range of force exertion (Range-F_{ST}) were assessed.

RESULTS

In condition I, ST produced 4.7% (5.1%) and 30.3% (18.3%) of its peak force (87.6N (30.5N) at KA=16.6° (17.9º)) in KA=120° and 90º, respectively (Figure 1). ANOVA (factors KA and condition) showed significant main effects of both factors, but no significant interaction. Condition II caused on average F_{ST} increase of 33.6%. However, condition III caused no further changes. Kruskal-Wallis tests showed no significant effect of conditions II and III on Range-F_{ST}. Therefore, only the first hypothesis was confirmed.

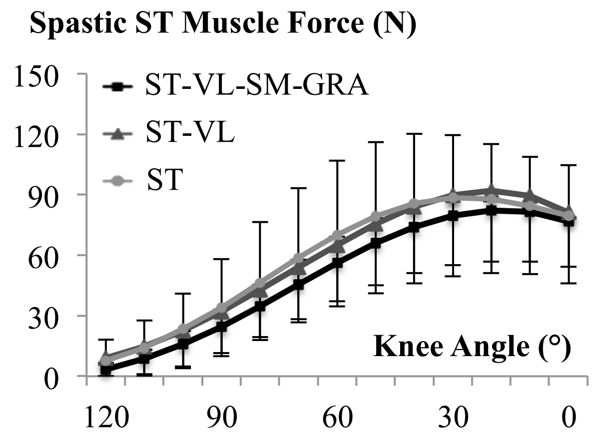

Figure 1. The knee angle-force (KA-F_{ST}) characteristics of semitendinosus

DISCUSSION

Elevated activated spastic ST forces due to inter-antagonistic EMFT are remarkable. Stabilizing the joint affected from spasticity by co-activating the antagonistic muscles is essential for the patients. However, due to inter-antagonistic EMFT, the effects of co-contraction may not be limited to a simple counter action on the joint, but also an enhanced agonistic torque is likely. Yet, patients with crouched gait already experience difficulty in extending their knee. Our data suggest that inter-antagonistic EMFT may play an adverse role in that. Unlike a previous study, which did report major narrowing of spastic GRA muscle's excursion due to co-activation of the vastus medialis [4], inter-antagonistic EMFT did not change Range-F_{ST}. Added inter-synergistic EMFT caused no further effects. Therefore, EMFT effects on spastic ST are limited to a sizable increase in its force production. Taking that into account and based on low force production in flexed knee positions and wide excursion of spastic ST shown, this study suggests that spastic ST may not be a dominant source of the pathological knee condition. Other spastic knee flexors should be tested for a broader understanding of their mechanics and effects of EMFT.

REFERENCES

1. Ates F. et al. (2013) Clin Biomech (Bristol, Avon), 28(1):48-54.
2. Ates F. et al. (2016) J Electromyogr Kinesiol, 28:130-6.
3. Pamuk U. (2016) J Mech Behav Biomed Mater, 63:207-19.
4. Ates F. et al. (2014) Clin Biomech (Bristol, Avon), 29(8):943-9.

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DISCLOSURE STATEMENT

The authors have no conflicts of interest to disclose.

Best Leg Length Discrepancy in Children with Spastic Hemiplegic Cerebral Palsy

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Introduction

The natural history of leg length discrepancy (LLD) is poorly understood, and consensus regarding the magnitude of discrepancy that merits surgical intervention does not exist for children with cerebral palsy (CP).[1-3] To maintain a functional gait pattern, individuals with a significant LLD will use compensatory mechanisms such as vaulting or circumduction.[2] As LLD increases, the compensatory mechanisms become more prominent, and the gait pattern becomes less efficient. This institution has identified kinematic ranges for normal gait pattern, based on averaging kinematic data from healthy individuals and considering one standard deviation away from those averages to be normal. This study aims to determine the maximum magnitude of LLD while maintaining the most efficient gait pattern.

Clinical Significance

Since consensus regarding the maximum magnitude of LLD while still maintaining the most efficient gait pattern does not exist, timing of surgical intervention and creation of treatment plans to maintain or treat LLD in the hemiplegic (HP) CP population becomes difficult. By having a set range of LLD that allows for the most efficient gait pattern, timing of surgical interventions and other treatment to achieve the appropriate LLD will be easier.

Methods

This retrospective study included 3-18 year-old subjects with spastic HP CP and LLD, a clinical leg-length measurement, a standing hip-to-ankle radiograph, scanogram, or CT leg-length measurement, and a gait analysis. An ICD-9 code search from 1995-2013 was conducted. Data was collected from clinical evaluation, gait lab analysis, and radiographs to acquire composite measurements of LLD and kinematic data of the patient's gait. Linear line regression was used to analyze correlations between LLD and gait variables.

Results

Data was obtained from 87 gait analyses (45 female, 42 male) (41 longer left leg, 36 longer right leg) for kinematic analysis. Initial findings from kinematic data revealed that to maintain normal pelvic obliquity and maximum hip flexion when the uninvolved leg is longer, an average LLD of less than 1.75cm ($p<0.05$) and 1.09cm ($p<0.05$), respectfully, is required. Other kinematic data values, such as knee flexion/extension and ankle dorsi/plantar flexion, which are used to identify compensatory mechanisms for LLD, were obtained; however there was no statistical significance. There was no statistical significance comparing a longer left or right leg. Initial analysis also observed a negative correlation between Gait Deviation Index (GDI) and magnitude of LLD, but this was not statistically significant ($p=0.48$). The average GDI ranged from

64.871.33, which is lower than reported ranges for GDI in the HP CP population (range 77.7-91.2).[4]

Figure 1

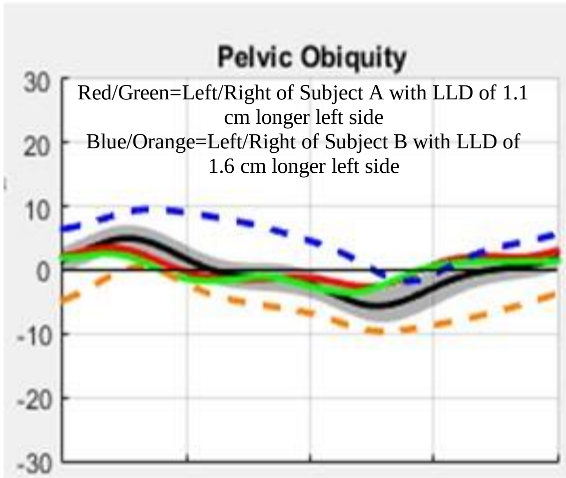

Figure 2

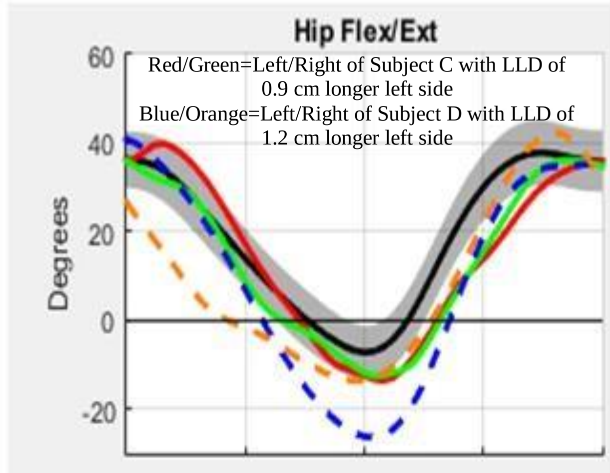

Figure 1 and 2 represent a comparison between the aged match normal kinematic ranges(black line represents the average value and gray area is one standard deviation away from the average), to kinematic values of subjects from the study, whose LLD is above and below the maximum LLD for normal kinematic throughout the gait cycle. Figure 1 represents Pelvic Obliquity, Figure 2 represents Hip Flexion/Extension.

Discussion

To maintain a normal pelvic obliquity, a maximum LLD of 1.75 is tolerated; however multiple kinematic variables, in addition to pelvic obliquity, contribute to a normal gait pattern. Initial findings show that as the magnitude of LLD increases, an individual's GDI decrease. With the small cohort it is not yet possible to determine the maximum magnitude of LLD tolerated before altered gait mechanics appear and the gait pattern deviates from being the most efficient. This cohort reveals no statistical significance comparing a longer left versus right leg. More data is needed before a definitive conclusion can be made.

References

1. Allen PE, Jenkinson A, Stephens MM, O'Brien T. Abnormalities in the uninvolved lower limb in children with spastic hemiplegia: the effect of actual and functional leg-length discrepancy. *Journal of pediatric orthopedics.* 2000;20(1):88-92.
2. Gurney B. Leg length discrepancy. *Gait & posture.* 2002;15(2):195-206.
3. Krick J, Murphy-Miller P, Zeger S, Wright E. Pattern of growth in children with cerebral palsy. *Journal of the American Dietetic Association.* 1996;96(7):680-685.
4. Rozumalski A, Schwartz MH. The GDI-Kinetic: a new index for quantifying kinetic deviations from normal gait. *Gait & posture.* 2011;33(4):730-732.

Disclosure Statement

The patient population, data, and analysis was all provided and completed at Children's Hospital Colorado. There were no conflicts of interest.

EFFECT OF SELECTIVE DORSAL RHIZOTOMY ON ANKLE JOINT STIFFNESS

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INTRODUCTION

Selective dorsal rhizotomy (SDR) is a neurosurgical technique used to reduce muscle spasticity of cerebral palsy (CP) patients. Even though improvement in patient mobility has been reported after SDR to last for many years, SDR has not prevented contracture formation [1]. Muscle contracture has been associated with muscle shortening, decreased joint range of motion, and increased joint stiffness. Spasticity induced enhanced joint stiffness has (i) a passive component due to changes in material properties of muscular and non-muscular structures and (ii) an active component due to neural impairment and changes in dynamic stiffness. SDR is used to interfere with the reflex circuit and, potentially, improve joint stiffness. Considering the recent report that SDR does not prevent muscle contractures, the effects of SDR on joint stiffness should be investigated to better understand muscle contracture formation following SDR. The main goal of this study was to test the hypothesis that SDR causes ankle joint dynamic stiffness of patients with CP to decrease.

CLINICAL SIGNIFICANCE

To understand muscle contracture formation after SDR and the use of advanced rehabilitation to improve mobility, it is essential to understand the short term effects of SDR on ankle joint stiffness.

METHODS

The study was approved by the Institutional Review Board of Mayo Clinic. 10 patients with CP (average of 6 years 1 month old at the time of surgery) who had previously undergone SDR were included. Patients were studied 3 months (range 1 - 8 months) before (Pre-SDR) and 1 year 1 month (range 10 - 17 months) after (Post-SDR) surgery. Clinical tests and gait analyses were performed. The Gross Motor Functional Classification System (GMFCS) was used to classify the mobility of the patients. Ankle passive range of motion (ROM) was recorded with the knee angle (KA) at 0º and 90º. 3D gait analysis and surface EMG (MA300, Motion Lab Sys, Inc., Baton Rouge, LA) were synchronized. The tonus of m. gastrocnemius was defined as one of the determinants of ankle joint stiffness. Therefore, its phasic character was scored using the dynamic EMG vs gait cycle data as follows: 0 = Phasic; 1 = Increased Phasic Activity; 2 = Continuous Low Level Activity; 3 = Continuous with Phasic Pattern; 4 = Continuous High Level Activity. The max plantar flexion (PF) and dorsiflexion (DF) angles during walking were calculated for each participant. Ankle joint dynamic stiffness was calculated as the slope of the ankle moment as a function of ankle angle during the second rocker of the gait cycle as described by Davis and DeLuca [2]. Separate paired student's t-tests were used to evaluate (i) if ankle joint passive ROM and (ii) max PF and DF angles during walking changed significantly after SDR, and (iii) if the changes in dynamic stiffness changed significantly after surgery as well as compared to previously published [2] ankle joint stiffness values collected from healthy participants. Dynamic EMG scores were assessed by using Wilcoxon signed-rank test. Differences were considered significant if $p<0.05$.

RESULTS

SDR improved the GMFCS level of half (5/10) of the patients. Ankle joint passive ROM measured at 0º of KA increased significantly ($p=0.01$) which indicated less contribution of m. gastrocnemius to passive joint stiffness after SDR. Whereas, no significant change in ankle ROM was observed at the 90º KA ($p=0.20$) which indicated no change on contribution of m. soleus on joint stiffness due to SDR. Dynamic EMG analysis showed that after SDR the phasic pattern scores of m. gastrocnemius decreased ($p<0.001$). SDR caused max PF during walking to decrease significantly ($p=0.01$) (Pre- and Post-SDR mean (SD) = -27.55º (18.54º) and -16.62º (12.53º), respectively) whereas it caused max DF to increase ($p=0.01$) (Pre- and Post-SDR mean (SD) = 2.69º (13.15º) and 10.94º (8.48º), respectively). Ankle joint dynamic stiffness also increased significantly ($p=0.01$) after SDR surgery (Figure 1). Therefore, our hypothesis was rejected. Moreover, the differences were not significant between Post-SDR values and the dynamic joint stiffness of healthy adults reported previously [2] ($p=0.43$).

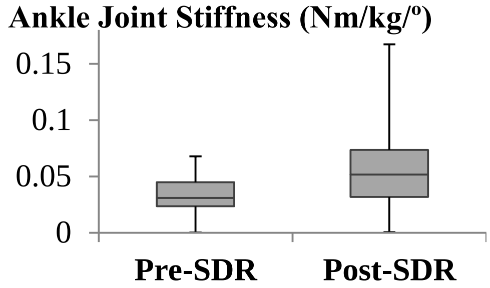

Figure 1. Box-and-whisker plot of ankle joint dynamic stiffness.

DISCUSSION

The present study showed that SDR promoted improved short term mobility for most of the patients. This is consistent with the previous reports [e.g. 1], It also improved passive ankle ROM as well as dynamic mobility parameters, e.g. a decrease in EMG pattern scores of m. gastrocnemius and an increase in DF angle during walking, thereby indicating a decrease in equinus gait. On the other hand, dynamic ankle joint stiffness increased after SDR. This may be one of the reasons that contracture formation is not stopped by SDR. Nonetheless, the dynamic ankle joint stiffness was close to the magnitude reported for healthy adults. Hence, the enhanced dynamic stiffness may indicate better motor control for patients with CP.

The passive tests showed improvement in m. gastrocnemius ROM (at 0º KA) but not for the soleus (at 90º KA). The m. gastrocnemius EMG revealed a more phasic dynamic pattern after SDR. We do not know the dynamic characteristic of the soleus EMG, but we have shown that dynamic joint stiffness which is the result of combined effects of m. gastrocnemius and soleus increased. Therefore, the soleus could be responsible for the increased dynamic stiffness. Additional studies are needed to identify both passive and active contributions of each muscle by using e.g. ultrasound elastography methods.

REFERENCES

1. Tedroff K. et al. (2011) Dev Med Child Neurol, 53:724-9.
2. Davis RB and DeLuca PA. (1996) Gait Posture, 4:224-31.

DISCLOSURE STATEMENT

The authors have no conflicts of interest to disclose.

COMPREHENSIVE LONG-TERM OUTCOMES 10-17 YEARS AFTER SELECTIVE DORSAL RHIZOTOMY

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INTRODUCTION

Selective dorsal rhizotomy (SDR) is a surgical method for reducing spasticity in children with cerebral palsy (CP). To date, there is no study with a well-matched control group examining whether SDR leads to lasting improvements in gait and function, or reduces the need for subsequent orthopedic intervention.

CLINICAL SIGNIFICANCE

Better understanding of comprehensive long-term outcomes following SDR is necessary to aid in treatment planning and counseling of patients who may be in need of spasticity management.

METHODS

Participants had spastic diplegic CP, completed a baseline gait analysis, underwent an SDR >8 years prior to enrollment, and were 16-25 years old at follow-up. Controls were matched on important clinical parameters at baseline, but did not undergo SDR. Barefoot kinematics, metabolic energy expenditure, and a physical exam were collected at baseline and follow-up. Participants completing a follow-up gait analysis were a subset of a larger cohort who completed survey measures on quality of life, pain, frequency of participation, satisfaction with life, function, and intervention history. Intervention details included lower extremity orthopedic surgery and botulinum toxin A (BTA) injection. Surgery on, or injection of each unique muscle/bone was counted as one instance. The Gait Deviation Index (GDI) was calculated as a measure of overall gait pathology [1].

RESULTS

The cohort consisted of 24 SDR and 11 control participants. Of these, 13 SDR (5m/8f, 17y8m (1y9m)) and 8 control participants (3m/5f, 19y7m (2y8m)) completed baseline and long term follow-up gait analysis. Ashworth scores for spasticity significantly decreased in SDR participants ($p<.05$) [Table 1]. The GDI improved more in controls than SDR ($\Delta_{control}=12$ vs. $\Delta_{SDR}=8$, $p<.01$), and the control group GDI was significantly higher than the SDR group GDI at follow-up ($GDI_{control} = 79$ vs. $GDI_{SDR} = 69$, $p<.01$) [Table 1]. Compared to the SDR group, control participants underwent significantly more soft tissue and bony orthopedic surgery and BTA injections (7.2 vs. 5.2 soft tissue, 6.4 vs. 5.5 bony, 19 vs. 7.5 BTA; all $p<.05$). Both groups reported low pain interference and high quality of life. There were no differences within survey responses.

Table 1. Summary of mean physical exam and gait-related measures for subset of participants that completed both baseline and follow-up gait analyses

	Baseline		**Follow-up**		**Change**	
	Control	**SDR**	**Control**	**SDR**	**Control**	**SDR**
Number	8	13	8	13		
Sex (Male/Female)	3/5	5/8	3/5	5/8		
Mean Age (SD)	5y 10m[a] (1y 0m)	4y 10m[a] (0y 10m)	19y7m (2y 8m)	17y 8m (1y 9m)	13y 9m*	12y 10m*
Spasticity (SD)						
Adductors	2.1[a] (0.7)	3.1[a] (0.8)	1.8[b] (0.9)	1.1[b] (0.3)	-0.3	-2.0*
Hamstrings	1.6 (0.5)	1.9 (0.7)	1.6[b] (0.6)	1.0[b] (0.1)	0.0	-0.9*
Hip Flexor	1.4 (0.5)	1.9 (0.8)	1.3[b] (0.4)	1.0[b] (0.0)	-0.1	-0.9*
Plantar Flexor	2.7 (0.4)	3.2 (0.7)	2.2[b] (0.7)	1.0[b] (0.0)	-0.5*	-2.2*
Rectus Femoris	1.8[a] (0.7)	2.9[a] (0.8)	1.8 (0.9)	1.2 (0.6)	-0.0	-1.7*
Range of Motion						
Gait						
GDI	67	61	79[b]	69[b]	12*	8*
Energy Cost [% speed matched normal]	267	312	206	242	-61	-70*

GDI=gait deviation index; [a] p<.05 for differences between groups at baseline, [b]p<.05 for differences between groups at follow-up; *p<.05 for baseline to follow-up changes within group;

DISCUSSION

SDR was an effective method to reduce spasticity long-term. Control participants had larger improvement in GDI, but underwent significantly more intervention in order to achieve these results. There were no differences between groups in survey measures. These data suggest that differing treatment course provides similar outcomes in early adulthood.

REFERENCES

1. Schwartz, M. et al. (2008) Gait Posture, 28(3):351-357, “in press”.

DISCLOSURE STATEMENT

Authors have no conflicts of interest to disclose.

HOW DOES HEMISPHERECTOMY AFFECT GAIT?

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INTRODUCTION

Hemispherectomy is an effective neuro-surgical intervention used for treatment of refractory epilepsy via disconnection of the hemisphere containing the seizure focus [1]. While many patients have a pre-operative motor impairment [2], the surgery inevitably results in a hemiparesis on the contralesional side [3]. Differences in surgical outcomes are influenced by several factors including seizure etiology, duration of epileptic disease prior to surgery, pre-operative development, and post-operative seizure outcome [3-8]. Some studies have examined changes in ambulatory status [7-9], but quantitative analysis of gait following hemispherectomy has not been described. The purpose of this study was to analyze instrumented gait data of children who had hemispherectomy to establish a baseline description of gait in this population. We hypothesized that gait function and quality would be lower and more variable in children post-hemispherectomy than typically developing children as measured by three-dimensional gait analysis.

CLINICAL SIGNIFICANCE

An understanding of changes in gait following hemispherectomy will enable physicians to more accurately understand the potential functional impact on the patient and counsel families regarding expected post-operative gait performance.

METHODS

Following IRB approval, a retrospective review was performed on children who underwent hemispherectomy and had an instrumented gait analysis before the age of 25 at Children's Hospital Colorado. Instrumented gait data including kinematic, kinetic, and temporospatial values were collected and gait deviation index (GDI) was calculated as standard of care between 2000 and 2016. Details regarding patients' seizure disorder and surgery were extracted from the medical record including seizure etiology, age of onset, duration of disease prior to surgery, type of procedure and laterality. Seizure etiology was classified as developmental, acquired, or progressive.

RESULTS

Thirty-four patients (21 female) satisfied the inclusion criteria. Mean age at the time of the hemispherectomy was 7.9±5.3 years old. Twenty patients had only post-operative gait analyses, and seven patients had both pre- and post-operative analyses. The remaining patients had only pre-operative gait data and were not included in this analysis.

Gait deviation index (GDI), a summary measure of kinematics, was calculated for all patients. Figure 1 shows that all but 2 subjects had post-operative GDIs below the average range of 90-100 (mean=100, SD=10). Four of the five patients who had multiple post-operative gait analyses demonstrated increases in GDI over time, though these differences were not significant. The fifth patient showed an initial increase followed by a decrease below baseline performance (Figure 1). The patients with both pre- and post-operative gait analyses showed variable results for changes in GDI, cadence, and step length when the pre-op analysis was compared with the initial post-op analysis. GDI increased in three patients and decreased in four patients, though these changes were not significant (Figure 2). Significant differences were not observed in the other gait quality or performance variables, and no differences were observed between the three seizure etiology classification groups.

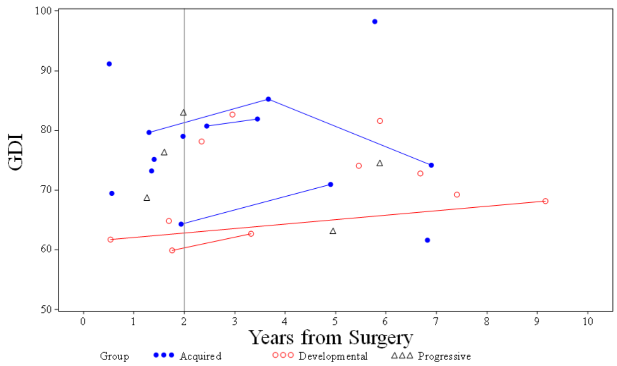

Figure 1: Post-operative GDI for acquired, developmental, and progressive etiologies. Lines represent individuals with more than one gait analysis

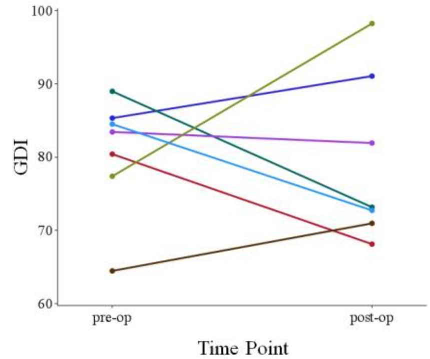

Figure 2: Individual changes in GDI from pre- to initial post-op gait analyses

DISCUSSION

These data support the hypothesis that children who undergo hemispherectomy experience lower and more variable gait performance than typically developing children. However, many patients in this retrospective analysis did not have pre-operative gait analyses to examine changes in gait performance. Those who had multiple post-operative analyses showed an upward trend in gait performance over time. To further investigate and better understand gait changes following hemispherectomy, a prospective study is necessary. As this is an uncommon procedure, a multi-site prospective study is warranted. Multidisciplinary collaboration, including the network of providers caring for children with intractable seizures and providers experienced in gait analysis, will facilitate this research and will provide a comprehensive, patient-centered approach for these children and families to optimize functional post-operative gait outcomes.

REFERENCES

1. Villemure, J.G., (2003) Epileptic Disord, 5(Suppl 2):S125-S130
2. van Empelen, R., et al. (2004) Brain, 127(Pt 9):2071-2079
3. Vining, E.P., et al. (1997) Pediatrics, 100(2 Pt 1):163-171
4. Devlin, A.M., et al. (2003) Brain, 126(Pt 3):556-566
5. Battaglia, D., et al. (1999) Neuropediatrics, 30(6)307-313
6. Wyllie, E. (1996) Epilepsia, 37 Suppl 1:22-25
7. Bode, S.d., et al. (2005) J Child Neurol, 20(1)64-75
8. Moosa, A.N., et al. (2013) Epilepsia, 54(10):1771-1779
9. Basheer, S.N., et al. (2007) Epilepsia, 48(1):133-140

DISCLOSURE STATEMENT

The authors have no conflicts of interest to disclose.

Podium Session #8

DATA METHODS/MODELING

MODERATED BY: **Adam Rozumalski, PhD:** Center for Gait and Motion Analysis
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Jason Long, PhD: Director, Motion Analysis Laboratory
Cincinnati Children's Hospital Medical Center, Cincinnati, OH

1. **The Ability of Accelerometry and Gyrometry of the Pelvis to Discriminate Two Different Balance Conditions**
   Antoine Brabants, Jim Richards, Kevin Deschamps, Jessie Janssen, Ambreen Chohan, Louise Connell

2. **Altered Gait Parameters Due to Restricted Ankle Function**
   Anahid Ebrahimi, Teresa Ferrara, Michael Christensen, Jill Higginson, Steven Stanhope

3. **Personalized Simulation of Joint Mechanics from Standard Motion Analysis and Clinical Imaging Data**
   Colin Smith, Darryl Thelen

4. **Soft Tissue Artifact Causes Significant Errors in The Calculation of Joint Angles and Range of Motion at the Hip**
   Niccolo Fiorentino, Penny Atkins, Michael Kutschke, Justine Goebel, K. Bo Foreman, Andrew Anderson

5. **The Smallest Number of Trials Needed to Reliably Measure Minimum Toe Clearance Variability During Level Ground Gait**
   Sylvester Carter; Mitchell Batavia, Gregory Gutierrez, Elizabeth Capezuti

6. **Simulation Modeling Analysis for Single-Subject Designs**
   James Richards, Kristen Nicholson

7. **Ensemble Average Vs Representative Cycle: How Should Gait Kinematics Be Presented?**
   Richard Pimentel, James Carollo

8. **A Comparison of Two Non-invasive Methods for Measuring Scapular Kinematics in Functional Positions**
   Elizabeth Rapp, R. Tyler Richardson, Stephanie Russo, William Rose, James Richards

THE ABILITY OF ACCELEROMETRY AND GYROMETRY OF THE PELVIS TO DISCRIMINATE TWO DIFFERENT BALANCE CONDITIONS

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INTRODUCTION

There are some quick, easy and cheap tools for assessing balance clinically [1,2]. However, these present several drawbacks including limited sensitivity to change, ceiling effects[3] and evaluator bias[4]. Pelvis acceleration has been shown to be a useful measure of postural stability [5-8] and has been demonstrated to be more sensitive compared to force plates[9]. However, the use of measures of angular velocity using gyroscopes have been largely ignored. This study aimed to assess the pelvis accelerations and angular velocity data to assess two different balance conditions with the view to identify the most relevant features of postural stability.

CLINICAL SIGNIFICANCE

Inertial Measurement Units (IMUs) which can measure acceleration and angular velocity are now beginning to be used in clinical assessment.

METHODS

We recruited 17 healthy individuals (9 males and 8 females) aged between 18 and 65. Participants were free from musculoskeletal injuries or neurologic disorders. Accelerometer and gyroscope data were collected at the pelvis using Delsys Inertial Motion sensors (Trigno™ System, Delsys Inc.). Participants were asked to perform a single-leg stance on a concrete floor with a thin carpet (Firm) and on an Airex Balance-pad (Foam). Each of these tasks were performed on the dominant leg and non-dominant leg. A two-way repeated measures ANOVA with pairwise comparisons was performed comparing surface and limb dominance.

RESULTS

Significant differences were seen between the two surfaces for the pelvis accelerations and angular velocity data in all three planes ($p<0.01$). Linear acceleration showed a 58%, 4%, 44% change in the three directions between firm and foam surface. The most sensitive measures in relation to percentage change and effect size were in the angular velocity data, with 166%, 280%, 363% changes in the sagittal, coronal and transverse planes of the pelvis respectively, table 1.

DISCUSSION

One possible explanation for the greater sensitivity is that the neuromuscular system is affecting rotational control and therefore having a direct effect on pelvis angular velocities with a secondary effect on linear accelerations.

Table 1: Pelvis IMU data				
Parameter	**Conditions**	**Mean (+/-SE)**	**CI of the differences**	**p value, effect size**
Medio-lateral acceleration (m/s)	Firm	0.183 (0.019)	0.030 to 0.184	$p=0.010$, $\eta_p^2=0.37$
	Foam	0.290 (0.032)		
	% Change	58.5%		
Vertical acceleration (m/s)	Firm	1.147 (0.010)	0.017 to 0.045	$p<0.001$, $\eta_p^2=0.59$
	Foam	1.178 (0.011)		
	% Change	2.7%		
Antero-posterior acceleration (m/s)	Firm	0.262 (0.034)	0.049 to 0.180	$p=0.002$, $\eta_p^2=0.48$
	Foam	0.377 (0.041)		
	% Change	43.9%		
Sagittal plane angular velocity (deg/s)	Firm	5.6 (0.7)	3.6 to 15.1	$p=0.004$, $\eta_p^2=0.44$
	Foam	15.0 (3.2)		
	% Change	166.6%		
Transverse plane angular velocity (deg/s)	Firm	8.2 (1.4)	13.6 to 32.3	$p<0.001$, $\eta_p^2=0.64$
	Foam	31.1 (5.5)		
	% Change	280.1%		
Coronal plane angular velocity (deg/s)	Firm	4.8 (0.3)	8.4 to 27.0	$p=0.001$, $\eta_p^2=0.52$
	Foam	22.6 (4.5)		
	% Change	363.6%		

REFERENCES

[1] M.M. Giorgetti, B.A. Harris, A. Jette, Reliability of clinical balance outcome measures in the elderly., Physiother. Res. Int. 3 (1998) 274–83. http://www.ncbi.nlm.nih.gov/pubmed/9859135 (accessed November 20, 2016).

[2] K. Berg, K.E. Norman, Functional assessment of balance and gait., Clin. Geriatr. Med. 12 (1996) 705–23. http://www.ncbi.nlm.nih.gov/pubmed/8890112 (accessed November 20, 2016).

[3] L. Blum, N. Korner-Bitensky, Usefulness of the Berg Balance Scale in stroke rehabilitation: a systematic review, Phys Ther. 88 (2008) 559–566. doi:10.2522/ptj.20070205.

[4] M. Mancini, F.B. Horak, The relevance of clinical balance assessment tools to differentiate balance deficits, Eur. J. Phys. Rehabil. Med. 46 (2010) 239–248. doi:0000.

[5] M. O'sullivan, C. Blake, C. Cunningham, G. Boyle, C. Finucane, Correlation of accelerometry with clinical balance tests in older fallers and non-fallers, Age Ageing. 38 (2009) 308–313. doi:10.1093/ageing/afp009.

[6] N.W. Saunders, P. Koutakis, A.D. Kloos, D.A. Kegelmeyer, J.D. Dicke, S.T. Devor, Reliability and Validity of a Wireless Accelerometer for the Assessment of Postural Sway, J Appl Biomech. 31 (2015) 159–163. doi:10.1123/jab.2014-0232.

[7] N.R. Heebner, J.S. Akins, S.M. Lephart, T.C. Sell, Reliability and validity of an accelerometry based measure of static and dynamic postural stability in healthy and active individuals, Gait Posture. 41 (2015) 535–539. doi:10.1016/j.gaitpost.2014.12.009.

[8] C. Neville, C. Ludlow, B. Rieger, Measuring postural stability with an inertial sensor: Validity and sensitivity, Med. Devices Evid. Res. 8 (2015) 447–455. doi:10.2147/MDER.S91719.

[9] R.E. Mayagoitia, J.C. Lötters, P.H. Veltink, H. Hermens, Standing balance evaluation using a triaxial accelerometer, Gait Posture. 16 (2002) 55–59. doi:10.1016/S0966-6362(01)00199-0.

DISCLOSURE STATEMENT

No authors have conflicts of interest to disclose

ALTERED GAIT PARAMETERS DUE TO RESTRICTED ANKLE FUNCTION

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INTRODUCTION

A primary goal for individuals with walking impairments is to walk safely and comfortably in their community 1 . Often individuals with impairment can achieve this goal by walking with a compensatory strategy compared to able-bodied individuals. In order to improve rehabilitation protocols and orthotic or prosthetic design, it is imperative that we understand how the limbs compensate with an impairment.

The muscle-tendon structures surrounding the ankle serve a primary function in safe and steady ambulation. Researchers have reported that restriction of the ankle joint may lead to compensations at the more proximal lower limb joints, such as increased bilateral hip power generation with unilateral ankle restriction 2 and increased knee work with bilateral ankle restriction . However, the compensatory strategies used by the limbs will differ based on the level of ankle restriction. The purpose of our study is to directly compare the compensatory strategies used by the limbs when subjected to unilateral and bilateral partial ankle restriction.

CLINICAL SIGNIFICANCE

The evaluation of lower limb joint kinetics and kinematics with unilateral and bilateral restricted ankle function will provide valuable insight into a possible mechanism used by humans to adapt when walking with partial ankle impairment.

METHODS

Nine healthy subjects (5F/4M, 4 10 years, 75. 1 .2 kg) were fitted for articulating AFOs locked at 90 degrees between the shank and foot for each limb. Subjects walked on an instrumented split belt treadmill (Bertec Corp., Columbus, OH) at 0. statures/s for ten minutes in three different conditions while motion (Motion Analysis Corp., Santa Rosa, CA) and force data were collected. The subjects walked (1) with standard shoes (Shoes), (2) with an AFO on the dominant limb (right for all subjects) (RiAFO), and ( ) with AFOs on both limbs (BiAFO).

Sagittal plane angle, moment, and power data using traditional inverse dynamics calculations 4 were analyzed for the ankle, knee, and hip joints for each subject. Loading response was defined as the initial phase of stance (0-12% of gait cycle) while we define terminal stance as the last phase of stance (including pre-swing) (1- 2% of gait cycle) 5 . Differences in metrics between conditions were compared using several repeated measures ANO As with an overall *p* value of 0.05. The Bonferroni correction was used for adjusting all post-hoc comparisons using SPSS software (IBM Corp., Armonk, NY).

RESULTS

Peak ankle plantarflexion angle was significantly reduced on the right limb wearing the AFO compared to Shoes (reduction of 14.2 º in RiAFO and 15.0 º in BiAFO) (*p* 0.05). On the left side, peak ankle power absorption during loading response and terminal stance and peak power generation in terminal stance significantly decreased in the BiAFO compared to Shoes (5 %, %, and 7% reduction, respectively) and RiAFO (55%, 40%, and %

reduction, respectively). On the right side, these values decreased for both the BiAFO (%, 40%, 27% reduction, respectively) and RiAFO (%, %, 27% reduction, respectively) compared to Shoes (*p* 0.05). Table 1 lists the significant peak knee and hip moments and powers in the three conditions. There were no significant differences in stride length.

T Peak joint moments (Nm/kg) and powers (/kg) for the three conditions of ankle restriction. *Italicized* variables occur during terminal stance, while others occur during loading response where L denotes the left limb and R denotes the right limb.

	p-value	Shoes		RiAFO		BiAFO	
		M	**S D**	**M**	**S D**	**M**	**S D**
L Hip Flexion Moment	*0.002*	*0.84[a]*	*0.22*	*0.90*	*0.21*	*0.97[a]*	*0.22*
L Hip Extension Moment	0.054	-0.9 [b]	0.2	-1.04[b]	0.22	-0.9	0.22
L Knee Extension Moment	0.01	0. 5	0.24	0. [c]	0.21	0.99[c]	0.12
L Knee Flexion Moment	*0.001*	*-0.12[a]*	*0.11*	*-0.13[c]*	*0.10*	*-0.19[a,c]*	*0.07*
L Knee Power absorption	0.00	-1.41[a]	0.55	-1.45[c]	0. 0	-1. [a,c]	0.40
R Hip Extension Moment	0.004	-0.94[a]	0.1	-0.97	0.1	-1.07[a]	0.20
R Knee Power absorption	0.005	-1.55[a]	0.5	-1. 1	0.4	-1.9 [a]	0. 9

[a]BiAFO and Shoes (*p* 0.05); [b]RiAFO and Shoes (*p* 0.05); [c]BiAFO and RiAFO (*p* 0.05)

DISCUSSION

In general, the primary peak changes in gait kinetics and kinematics occurred during the loading response and terminal stance phases of gait. Restricting positive right ankle power by 27% compared to Shoes resulted in an increased hip extension moment on the left side during loading response. Restricting positive ankle power by 27% on the right and 7% on the left led to a subsequent increase in contralateral knee power absorption in loading response. It appears that these healthy individuals increased their left (non-dominant) hip flexion moment in terminal stance with a subsequent increase in right (dominant) hip extension moment in loading response. This was not observed while the left limb was in loading response, which instead compensated with an increased left ankle dorsiflexion moment. Interestingly, hip extension moment increased on the left during right ankle restriction, but increased on the right during bilateral ankle restriction. The different compensatory strategies used by the non-dominant and dominant limb with bilateral ankle restriction highlight the value in considering limb dominance during bilateral ankle impairment rehabilitation.

REFERENCES

1. Andrews, A. et al. (2010) ournal of geriatric physical therapy, ( ): 12 -1 4.
2. utzke, C. et al. (2012) ournal of Biomechanics, 45(14): 2405-2410.
. Huang, T. et al. (2015) ournal of Experimental Biology, 21 (22), 541- 550.
4. inter, D. (2009) Biomechanics and Motor Control of Human Movement, 4th ed.
5. Perry, . (1992) Gait Analysis: Normal and Pathological Function, 2nd ed.

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Personalized Simulation of Joint Mechanics from Standard Motion Analysis and Clinical Imaging Data

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INTRODUCTION

For several decades, motion analysis has played an important role in the diagnosis, treatment planning, and understanding of movement disorders. Despite its widespread use, the clinical application of this technology has remained relatively unchanged since its introduction. While the accuracy of motion analysis has reached its inherent limit due to skin motion artefact, opportunity exists to gain further insights through personalization of the underlying model and advanced simulation techniques.

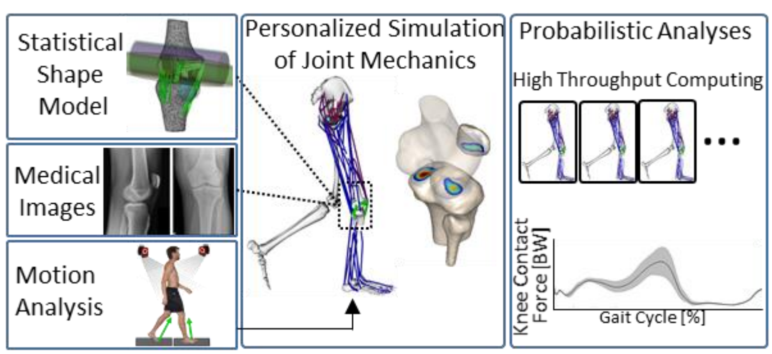
We utilized three recent advances in human movement research to develop a novel simulation framework which integrates standard motion analysis measurements with clinical imaging to predict joint mechanics during functional movement. First, a statistical shape model and medical images are used to morph a musculoskeletal model with a detailed knee joint to generate patient specific ligament and articular surface geometries. Second, a novel simulation routine uses the model and motion analysis data to simulate muscle forces, secondary joint kinematics, ligament forces and articular contact loading during a measured movement. Third, high throughput computing is used to perform probabilistic analyses (e.g. Monte Carlo) to quantify the uncertainty in simulation predictions. We demonstrate the functionality of this simulation framework by predicting muscle, ligament and cartilage loading at the knee joint during walking.

CLINICAL SIGNIFICANCE

By integrating standard motion analysis, clinical imaging and advanced musculoskeletal modeling and simulation techniques, further insights can be gained into joint mechanics during functional movement.

METHODS

A simulation framework was developed to personalize the geometries of a multibody knee model, simulate subject-specific joint mechanics and quantify the uncertainty in predicted quantities. A multibody statistical shape model of the knee which includes bone, cartilage and ligament attachment geometries was developed from segmented magnetic resonance images of 20 young healthy subjects. The model geometries can be morphed, enabling future personalization of the model to clinical imaging data. The model has 6 degree of freedom tibiofemoral and patellofemoral joints with explicit representation of ligaments and articular contact. The knee model has been integrated into a generic lower extremity musculoskeletal

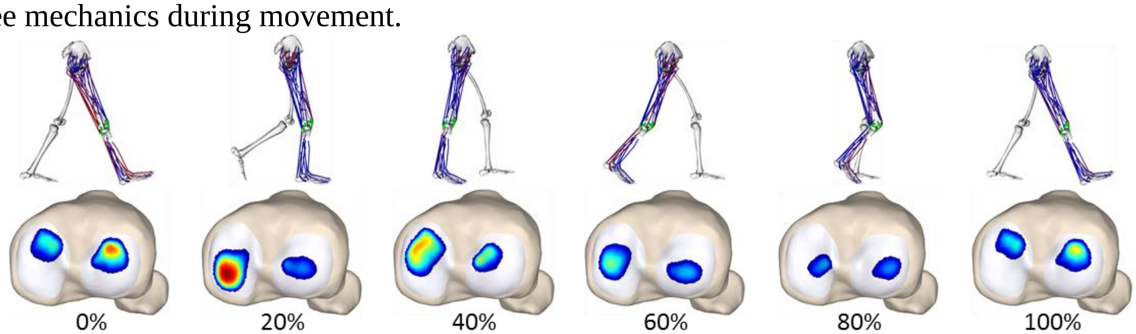


model and validated by comparing simulated knee kinematics against measured kinematics from dynamic MRI [1].

To simulate knee joint mechanics during walking, we developed a novel simulation routine, termed COMAK (concurrent optimization of muscle activations and kinematics), that predicts internal knee tissue loads from kinematic and kinetic measurements of movement. At each time step, COMAK calculates the muscle forces, secondary knee kinematics, ligament forces and cartilage contact pressures that minimize the weighted sum of squared muscle activations while generating the observed movement [2]. The uncertainty in simulation predictions due to model parameters, geometries, and muscle redundancy are quantified using Monte Carlo analyses. These probabilistic analyses are readily performed using a high throughput computing grid which enables thousands of simulations to be performed in parallel.

DEMONSTRATION

To investigate the influence of neuromuscular coordination and structural joint properties on knee behavior during walking, we used the high throughput computing cluster to perform probabilistic analyses. By treating model and simulation parameters as stochastic variables, we have assessed the sensitivity of predicted knee mechanics to neuromuscular coordination strategies and model geometric and constitutive properties. Recently, we have begun evaluating the use of statistical shape modeling to rapidly generate subject-specific knee models, as well as to study the influence of articular surface geometry and ligament attachment location on knee mechanics during movement.



SUMMARY

We introduce a novel simulation framework to predict personalized joint mechanics during functional movement from standard gait analysis and medical image data. Statistical shape modeling is used to personalize the geometries of a multibody knee model, the COMAK simulation algorithm is used to simulate knee joint mechanics during walking and high throughput computing is used to quantify the uncertainty in simulation predictions. We envision our novel approach will enable the next generation of motion analysis tools, in which subject-specific joint behavior can be simulated, and the effects of interventions can be predicted and considered for clinical treatment planning. The model and simulation routine are being implemented in OpenSim and will be made publically available through simtk.org.

REFERENCES

1. Lenhart RL, et al. (2015) Annals of Biomedical Engineering, 43(11): 2675-85.
2. Smith, CR, et al (2016). Journal of Knee Surgery 29(2): 99-106

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DISCLOSURE STATEMENT

Colin Smith and Darryl Thelen have no conflicts of interest to disclose.

SOFT TISSUE ARTIFACT CAUSES SIGNIFICANT ERRORS IN THE CALCULATION OF JOINT ANGLES AND RANGE OF MOTION AT THE HIP

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INTRODUCTION

Joint angles provide insights into movement abnormalities for clinical gait analysis and scientific investigations [1]. Most often, tracking of markers adhered to the skin surface serve as the basis for calculating joint angles, but skin marker motion capture suffers from soft tissue artifact (STA) [2]. Many previous studies of the lower limb used pins implanted into bone as a reference standard to quantify errors associated with STA, but this approach is highly invasive and may interfere with the movement of soft tissue layers between the skin surface and bone. In addition, STA has yet to be quantified for the hip joint during dynamic activities.

Therefore, the purposes of this study were to: 1) measure STA during dynamic activities that engage hip motion in each anatomical plane and 2) quantify errors in hip joint angle and range of motion measurements.

CLINICAL SIGNIFICANCE

Given the ubiquitous nature of skin marker motion capture in clinical gait analysis, the results of this study are important for clinicians and biomechanists who rely on joint angles calculated using markers adhered to the skin for diagnosis and treatment of pathologies.

METHODS

Eleven subjects signed informed consent to participate in this University of Utah IRB approved study (mean (SD) age: 23.2 (2.2) years, BMI: 21.1 (1.9) kg/m^2). Subjects were imaged simultaneously with a 10-camera Vicon motion capture system and high-speed dual fluoroscopy (DF), both at 100 Hz. Subjects performed six activities: standing (static), level walk, incline (5°) walk, abduction (45°), internal hip rotation and external hip rotation. Spherical markers (14mm diameter) were placed bilaterally on the pelvis at the anterior superior iliac spines (ASIS), posterior superior iliac spines (PSIS) and lateral aspect of the iliac crest (ILC) [3]. Markers were placed on the femur at the greater trochanter (GRT), thigh (rigid cluster of four) (THI) and lateral knee (KNE) [3]. Bony landmark positions were determined from DF images using model-based tracking [4], a technique our lab validated to a bias and precision of less than 1mm and 1° in the hip [5]. Skin marker positions were transformed into the DF coordinate system via an acquisition of custom skin markers containing metal beads [6].

STA was defined as the range of skin marker positions in the DF-measured bone segment axes. To minimize the effects of anatomical variability, joint angles were offset by their value during the static activity. Joint angle and range of motion (ROM) errors were defined as the value measured from skin markers (SM) relative to the value measured from DF (i.e., SM – DF). A six degrees-of-freedom model in Visual3D was used to calculate joint angles. Data were reported as mean (95% confidence interval) across all subjects and activities. Statistical comparisons were made using a multivariable linear regression model in Stata.

RESULTS

When considering all activities, STA varied from 0.4 (0.3 0.5) cm for the PSIS marker to 3.1 (2.6 3.6) cm for the GRT marker, both in the anterior-posterior direction (Fig. 1). STA demonstrated a dependence on anatomical direction as well as marker location (Fig. 1).

Hip joint angle and range of motion measurements were found to differ between skin marker and dual fluoroscopy measurements (Table 1). Skin marker measurements found the hip more extended (i.e., less flexed), more adducted and more internally rotated. For all rotation directions, skin markers generated smaller ROMs than dual fluoroscopy.

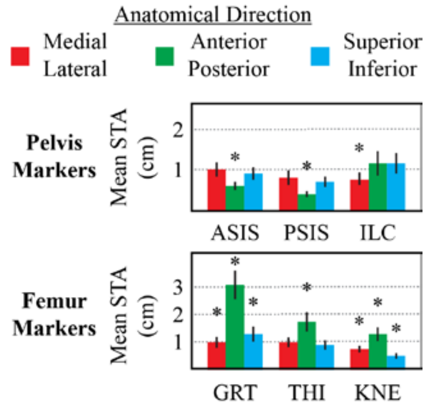

Figure 1: STA during all dynamic activities.
* P < 0.05 vs. other anatomical directions

Table 1: Hip joint angle and range of motion errors during all dynamic activities.

	Flexion(+) Extension(-)	Abduction(+) Adduction(-)	External (+) Internal(-) Rotation
Joint angle (°)	-1.9 (-2.5 -1.2)*	-0.6 (-1.0 -0.3)*	-6.0 (-7.5 -4.5)*
Range of motion (°)	-4.1 (-4.9 -3.4)*	-1.8 (-2.7 -1.0)*	-7.7 (-10.3 -5.0)*

* P < 0.05

DISCUSSION

To our knowledge this study represents the first assessment of STA at the hip during dynamic activities. STA magnitude for femur markers was similar to, or larger than, those reported in a review of the literature [2]. The reduced joint angle and range of motion measurements for skin markers support the “skin sliding” theory [2] that posits skin can slide inharmoniously relative to the underlying bones. Specifically, results presented herein suggest the femur and pelvis move beyond the change in position measured with skin markers. These results motivate future work to minimize the errors imparted by STA at the hip.

REFERENCES

1. Wren, T.A. et al. (2011) Gait & Posture 34(2):149-153.
2. Leardini, A. (2005) Gait & Posture 21(2):212-225.
3. Fiorentino, N.M. et al. (2016) Gait & Posture 50:246-251.
4. Bey, M.J. et al. (2006) Journal of Biomechanical Engineering 128(4):604-609.
5. Kapron, A.L. et al. (2014) Journal of Applied Biomechanics 30(3):461-470.
6. Fiorentino, N.M. et al. (2016) Annals of Biomedical Engineering 44(7):2168-2180.

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DISCLOSURE STATEMENT

The authors have no conflicts of interest to disclose.

THE SMALLEST NUMBER OF TRIALS NEEDED TO RELIABLY MEASURE MINIMUM TOE CLEARANCE VARIABILITY DURING LEVEL GROUND GAIT

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INTRODUCTION

Within-person Minimum Toe Clearance (MTC) variability of an individual's MTC distribution has been investigated for its potential relationship to trip-related falling. Age-related differences in within-person MTC variability have only been demonstrated in treadmill studies[1] but not in 3 overground studies.[2-4] However, the treadmill variability studies may produce erroneous results[5] especially in older adults who may not acclimate readily to this type of walking.[6] In contrast, the overground studies may more accurately replicate natural walking, but these studies may not have collected sufficient trials to obtain reliable MTC variability data. In the only study found to estimate the reliability of MTC variability data, Konig et al.[7] suggested that a minimum of 50 MTC values may be needed to reliably measure MTC variability for overground walking studies. However, Konig et al.[7] only evaluated young adults with MTC variability measured as the standard deviation (SD). It is unknown whether this reliability estimate may be different for older adults and other measures of MTC variability such as the interquartile range (IQR) or coefficient of variation (CV) that were evaluated in the 3 overground studies. Therefore, the present study was conducted to determine the minimum number of MTC values needed to reliably measure MTC variability (IQR, SD, CV) for young and older adults.

CLINICAL SIGNIFICANCE

Age-related differences in MTC variability may be associated with older adults' increased trip risk. Reliable MTC variability data is needed to accurately estimate this risk and reduce the potential for type II errors.

METHODS

Participants were 20 older (7 males, 13 females; mean age = 71.3 ± 7.2 years) and 20 young adults (9 males, 11 females; mean age = 28.8 ± 5.7 years) who were screened for factors affecting walking ability and fall risk. A modified Cleveland Clinic marker arrangement was used, and movement capture was via 6 Oqus cameras recording at 120Hz. All participants walked for 50 trials at their self-selected speed over a 7-meter walkway, resting 5-minutes after 25 trials. Test-retest reliability analyses were performed on 50 MTC values (25 left and 25 right) collected before and after the 5-minute rest. To prevent order effects, the order of the 50 MTC values was partially randomized by randomly choosing the first MTC value from the 50, then randomly selecting the second MTC value from those remaining, and so on until all 50 were chosen. Acceptable reliability was defined as an Intraclass correlation (ICC (2,1)) coefficient equaling or exceeding 0.9. Reliability was assessed with an Intraclass correlation (ICC (2,1)) for absolute agreement and the Standard Error of Measurement (SEM).

RESULTS

The number of MTC values needed to reliably ($r \geq 0.9$) measure MTC variability (IQR, SD, and CV) was 28, 45, and 49 for older adults and 45, 18, 16 for young adults, respectively (Figure 1). The SEM estimates at an acceptable reliability for MTC variability (IQR, SD, and CV) was 0.21, 0.11, and 0.06 cm for older adults and 0.15, 0.09, and 0.05 cm for young adults, respectively (Figure 2).

Figure 1: The number of MTC values needed to reliably measure MTC variability based on the Intraclass Correlation Coefficient (ICC) for A) IQR, B) SD, and C) CV data.

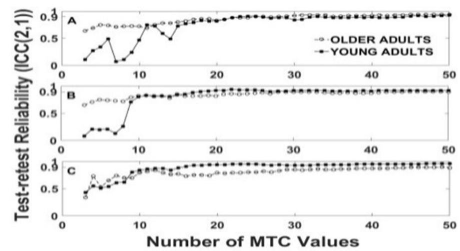
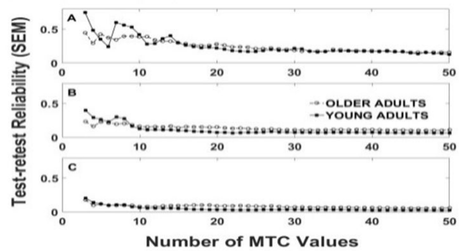


Figure 2: The number of MTC values needed to reliably measure MTC variability based on the Standard Error of Measurement (SEM) for A) IQR, B) SD, and C) CV data.



DISCUSSION

This study examined if age or different variability measures (IQR, SD, CV) affected the number of MTC values needed to reliably measure MTC variability. There was not a consistent pattern regarding which age group (older or young) achieved acceptable reliability with lower numbers of MTC values for each of the measures of variability suggesting that reliability should be evaluated in each age group. An acceptable reliability ($r \geq 0.9$) for measurements of MTC variability either as the IQR or SD was achieved with 45 MTC when both young and older adults were considered. However, MTC variability measured as the CV required 49 MTC values for older adults. The SEM, at an acceptable reliability level, was smaller for the CV followed by the SD and IQR for both young and older adults. Based on this present study and that by Konig et al.,[7] 2 of the previous overground studies[2, 4] may not have used sufficient MTC values to reliably estimate MTC variability. Although the third study [3] used 50 MTC values, because motion capture was by inertial sensors which are associated with larger errors than from optical motion capture systems[8] used in this present study and that by Konig et al.,[7] extrapolations may not be valid. Overground gait studies examining age differences in MTC variability (IQR, SD, CV) may require a minimum of 50 MTC values to ensure reliable estimates for both older and young adults.

REFERENCES

[1] Barrett RS, et al. (2010) Gait and Posture, 32:429-435.
[2] Garman CR, et al. (2015) Journal of Biomechanics, 48:1229-1232.
[3] Hamacher D, et al. (2016) Experimental Brain Research.
[4] Winter DA, et al. (1990) Physical Therapy, 70:340-347.
[5] Dingwell JB, et al. (2001) Journal of Biomechanical Engineering, 123:27-32.
[6] Wass E, et al. (2005) Gait and Posture, 21:72-79.
[7] Konig N, et al. (2014) Gait and Posture, 39:615-617.
[8] Kanzler CM, et al. (2015) Conference proceedings: IEEE Engineering in Medicine and Biology Society, 2015:5424-5427.

DISCLOSURE STATEMENT

We the authors declare that we do not stand to benefit financially from this publication.

Simulation Modeling Analysis for Single-Subject Designs

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INTRODUCTION

Repeated-measures experiments performed on groups cannot be used to objectively assess individual treatment benefit. Consequently, single subject analyses are gaining traction in medical research. Single subject methodology is founded in baseline logic: a measurable change in subject behavior following an intervention is probably due to the intervention (A-B design). If the intervention can be removed and re-introduced (A-B-A-B design), a stronger case can be made for the intervention as the reason for change.

While numerous statistical approaches can be used to analyze single-subject data, most suffer from the inability to address both of two primary issues that complicate single-subject analyses: autocorrelation and trend. In addition, most approaches are challenging to implement and/or require a large number of observations for each subject.

Simulation modeling analysis (SMA) offers a relatively new approach to analyzing time-series single-subject data that can be performed on a small number of baseline and post-intervention observations [1]. Unlike some approaches, it does not depend on normally distributed data and provides a measure of effect size. The approach provides good Type-I error control and adequate power with small time-series samples using the A-B design [2].

SMA provides an assessment of changes in both level (mean differences) and trend (slope differences). Baseline measures with low variability make it easier to detect measurable changes in level between phases, and for measures that are susceptible to learning or practice effects, changes in level should be considered in the context of trends. Figure 1 illustrates how trend can influence the interpretation of level effects.

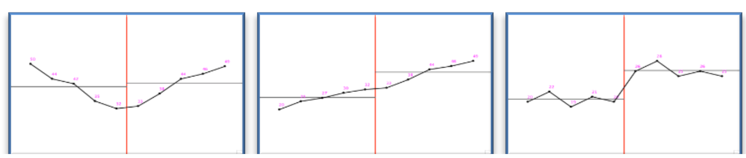

Figure 1. Three common trend results. Horizontal lines indicate phase means, vertical red line separates phases. The left plot shows change in trend but not level, center plot shows change in level confounded by trend, and right plot shows change in level with no trend.

CLINICAL SIGNIFICANCE

SMA for single-subject data provides an objective means of determining whether a treatment resulted in a measurable change in behavior on an individual basis. In addition, the ability to identify individual responses to treatment provides opportunity for further exploration into commonalities within both responsive and non-responsive groups.

METHODS

SMA requires approximately 5 baseline-phase measures (A) and 5 post-intervention measures (phase B). Phase A is coded as 0, and phase B is coded as 1 (Figure 2). SMA calculates the autocorrelation for the measured data, corrects for sample size, and generates approximately 5000 pseudo-random data sets with the same autocorrelation and number of observations as the original data. This is referred to as a modified “bootstrapping” process. Next, it calculates the correlation between the random data and coded phase data. The p-value is the proportion of randomly generated correlations that exceed the correlation of the original data.

DEMONSTRATION

For illustration purposes, pre-surgical (Pre) and post-surgical (Post) Gait Deviation Indices (GDIs) were calculated for 13 CP patients in two ways: 1) using all gait cycles from each patient pre- and post-surgery, resulting in two GDIs/patient, and 2) using each of 5 trials from each patient pre- and post-surgery, resulting in 10 GDIs/patient. Fifteen different surgical proceduress were represented in the data, with patients typically undergoing single event multilevel surgery. Pre/Post GDI values calculated from all gait cycles were analyzed for the group using a dependent t-test, while Pre/Post GDI values for individual trials were analyzed using SMA.

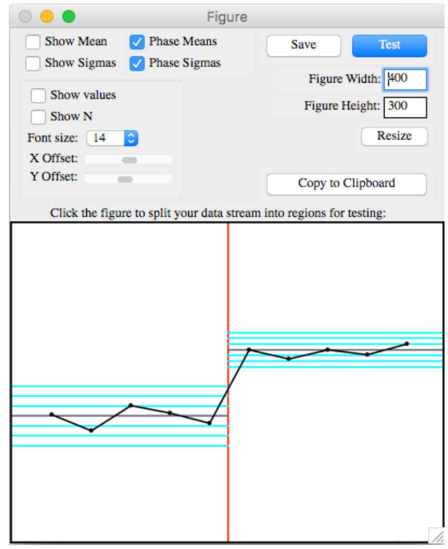

Figure 2. SMA setup screen. Points to the left of the red line are pretest (phase 0), and points to the right are posttest (phase 1). Blue lines represent standard deviation values.

Results of the t-test indicated that the Post GDI mean (76.5$\pm$7.9) was significantly higher than the Pre GDI mean (71.6$\pm$10.7), t=-2.242, p=0.045. From this result, we can conclude that surgical intervention had a positive effect on GDI scores, which reflects an overall improvement in the “normalcy” of gait among patients in the sample.

The SMA analysis (Table 1) indicates that 8 of the 13 patients demonstrated a significant increase in GDI score, while 2 patients demonstrated a significant decrease in scores, and 3 showed no significant change.

Table 1. Individual results from the SMA analysis. Green: significant increase in Pre/Post GDI, Red: significant decrease in Pre/PostGDI

Subject	R	Level (p)
Subj 1	0.977	0.0001
Subj 2	0.992	0.0001
Subj 3	0.972	0.0001
Subj 4	0.839	0.006
Subj 5	0.904	0.0004
Subj 6	-0.247	0.5638
Subj 7	0.951	0.0002
Subj 8	0.657	0.0072
Subj 9	0.966	0.0001
Subj 10	0.21	0.5412
Subj 11	-0.757	0.007
Subj 12	-0.666	0.0024
Subj 13	0.122	0.6062

SUMMARY

While traditional inferential statistics (ie. t-test) can determine group tendencies, SMA provides the ability to test single case experiments or to “dissect” group results. In the example above, it’s clear that a majority of cases improved. However, SMA determined that this result represented only 2/3 of the cases. SMA software is free online at: clinicalresearcher.org/software.htm.

REFERENCES

1. Borckardt, J. et al. (2014) Neuropsychological Rehabilitation, 24(3-4): 492-506.
2. Borckardt, J. et al. (2008) American Psychologist, 63(2): 77-95.

DISCLOSURE STATEMENT

The authors have no conflicts of interest to disclose.

**ENSEMBLE AVERAGE VS REPRESENTATIVE CYCLE:
HOW SHOULD GAIT KINEMATICS BE PRESENTED?**

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INTRODUCTION

Motion data from instrumented gait analysis (IGA) are typically portrayed in kinematic plots to describe normalized joint angles and general walking ability of the subject. Given the amount and complexity of IGA data, reports typically contain information from either a single representative cycle (RC) or an ensemble average (EA) of the cycles; although no clear consensus exists in the clinical gait community. During an IGA, multiple walking trials are collected and an EA of all valid gait cycles is computed. Using the EA, a single RC that closely resembles the EA can also be selected for use in clinical interpretation and decision making.

There are advantages and disadvantages to both techniques. The benefits of using EA are: a) it contains summed information of all the gait cycles present during various trials; b) the gait variability measurements provide additional information about the subject's gait [1]; and c) the averaged waveform may be more representative than a single cycle [1]. The benefits of using the RC are: a) the curve comes from one intact, definitive gait cycle; b) gait features (minima & maxima) are unchanged, as these features may be attenuated during EA; c) the waveform is not affected by any other gait cycles. These two techniques are mutually exclusive; the benefits of one are the drawbacks of the other. While it has been said that the EA forms the basis of clinical assessments [1], a carefully-selected RC may also serve that purpose.

Our purpose was to compare RC and EA representations of gait data, in terms of the kinematic root mean square (RMS) error between methods, and evaluate any differences detected may affect gait quality as measured by Gait Deviation Index (GDI) [2]. These were assessed in a group with cerebral palsy (CP) and an age-matched normal group (AMN).

CLINICAL SIGNIFICANCE

Representative gait cycles or ensemble averages of all collected data are legitimate ways to portray a person's motion data. It is important to understand the pros and cons of each data presentation method before making clinical decisions from either technique.

METHODS

Thirty-two participants were included in this analysis, 16 with CP and 16 AMNs. Mean characteristics for the group include (AMN vs CP): *age* 13.7 (4.8) vs. 13.8 (4.8) years; *height* 156.9 (16.7) vs. 152.8 (16.1) cm; and *weight* 49.2 (20.5) vs. 44.6 (14.7) kg (*Mean (SD))*. Participants began standing and walked through the capture volume at a self-selected pace. The first stride on each side was excluded to avoid the influence of gait initiation. An average of 14 valid strides were collected over at least 4 walking trials for each participant. Kinematics were collected using a lower body marker-set conforming to the Conventional Gait Model and captured in Vicon Nexus (v 2.2.3, Oxford, UK) [3]. Using Matlab (v R2015A, Mathworks, Natick, MA, USA), kinematics were parsed from gait events and interpolated to 2% increments of the gait cycle. Lower body kinematics only included the 9 curves necessary to calculate GDI

[2]. In order for the RC to contain an actual intact gait cycle, RCs were selected primarily on the side with lower GDI and then the contralateral side was chosen from the cycle directly before or after the ipsilateral cycle closest to the EA. Each subject had only one RC and EA across the multiple trials collected. RMS error was the absolute error between RC and EA waveforms, and GDI difference was calculated as EA – RC. Analysis of variance was used to test for significance of the EA-RC difference between the two groups. Statistical significance was set at α = 0.05.

DEMONSTRATION

There were no significant differences between the CP and AMN groups in age, height or weight (*P* ≥ .468). Average RMS error between EA and RC were similar between groups (Figure 1). At the whole group level, RMS error was also similar across body segments (Figure 2) and planes (sagittal, frontal, and transverse; *P* = .155). GDI scores between EA and RC were similar between groups for both sides (Figure 1).

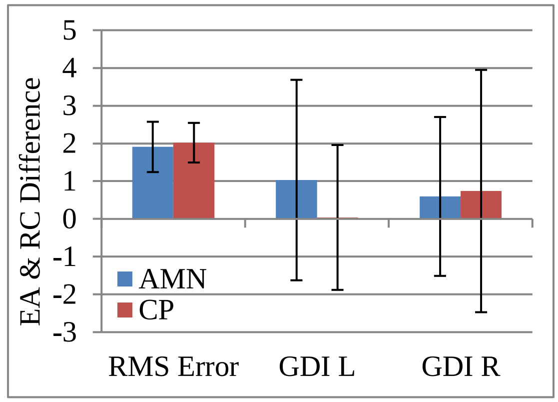

Figure 1: EA and RC are similar between groups (*P* ≥ 0.236). RMS units are degrees, while GDI is unit-less (score based).

SUMMARY

The purpose was to determine if there are differences between EA and RC data representation techniques. EA and RC are similar in accuracy to approximately 2° on average (0.5 SD). Overall gait quality is preserved in both techniques (GDI difference of approximately 1 score (2.5 SD)). In summary, kinematic gait data from IGAs can be accurately depicted using either data presentation method.

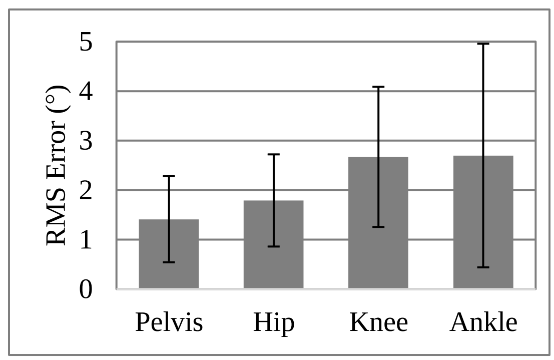

Figure 2: RMS error between segments across the whole group. None of the body segments were significantly different from the others (*P* = 0.161).

REFERENCES

1. Winter DA. 2009. Biomechanics and Motor Control of Human Movement, Fourth Edi. John Wiley & Sons Inc.
2. Schwartz MH et al. 2008. Gait Posture. 28:351–357.
3. Vicon. 2010. Plug-in Gait Product Guide — Notes.

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A COMPARISON OF TWO NON-INVASIVE METHODS FOR MEASURING SCAPULAR KINEMATICS IN FUNCTIONAL POSITIONS

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INTRODUCTION

Motion of the scapula is an essential consideration in the evaluation of upper extremity function, as abnormal scapular kinematics are associated with a variety of shoulder pathologies [1]. Traditional surface marker approaches are not applicable to the measurement of scapular motion, as the scapula translates beneath the skin. Several dynamic measurement techniques have been proposed with varying degrees of accuracy. The most widely utilized dynamic measurement approach is the acromion marker cluster (AMC); however, this method can produce errors in extreme humeral elevation [2] and in certain populations [3].

An alternative approach, based on an individualized linear regression, was recently proposed and validated in nine healthy adults, yielding root mean square (RMS) errors below 8° for all axes of scapulothoracic (ST) motion [4]. This approach utilizes calibration positions to define a subject-specific relationship between humerothoracic (HT) motion, acromion process (AP) position and ST orientation. This relationship is then used to estimate ST orientation in motion using the more easily measured HT and AP inputs.

The AMC and the regression approaches, while both suitable candidates for non-invasive measurement of scapular kinematics, have not been compared to each other, nor assessed for use in an adolescent population. This study evaluated the two approaches in five functional positions, comparing each method of estimation to palpated ST orientations. We hypothesized that the regression approach would outperform the AMC.

CLINICAL SIGNIFICANCE

The results of this study will inform future research as to the best scapular kinematic measurement technique for identifying dysfunction or evaluating efficacy of intervention.

METHODS

Eighteen healthy adolescents were recruited. Markers were placed on the trunk (T1, T8, sternal notch) and humerus (medial/lateral epicondyles), and an AMC was placed with the central marker directly on the AP. Subjects sat on a stool and moved through fifteen positions: neutral, abduction, external/internal rotation, flexion/extension, forward reach, hand to mouth, hand to nape, hand to spine, hand to contralateral shoulder, hand to back pocket, lateral reach, unconstrained humeral elevation and 90° scapular plane abduction. In each position, scapular landmarks were palpated (trigonum spinae and inferior angle) and markers were affixed.

The double calibration approach [5] was implemented for the AMC, interpolating the transformations from the neutral and abduction positions based on HT elevation angle. The regression approach employed a multiple linear regression on data from the first ten positions to generate three equations (one for each axis) that estimated ST angles from the HT and AP inputs. The last five positions were solely used as test positions, comparing the AMC and regression estimates to ST orientations determined by palpation. Accuracy was evaluated by both RMS errors and a three-way ANOVA (measurement method, position, axis of ST motion).

RESULTS

The regression approach yielded smaller RMS errors for every position across all axes of ST motion (Figure 1). The regression significantly underestimated angles in the hand to contralateral shoulder position (Mean difference = 1.8°, $p = .017$). The AMC estimated significantly less internal rotation than palpation (Mean difference = 4.8°, $p < .001$) (Figure 2).

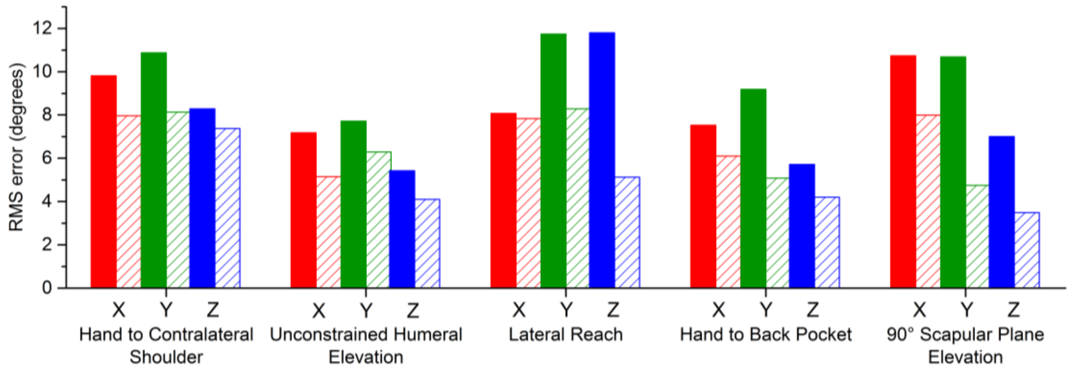

Figure 1: RMS errors in each position for each axis of ST motion. Solid bars are AMC errors, striped are regression errors.

DISCUSSION

The regression approach consistently produced smaller RMS errors for every position and every axis. Still, in the hand to contralateral shoulder position, this technique underestimated ST angles (primarily upward rotation and posterior tilt). For most subjects, the hand to contralateral shoulder position required more HT internal rotation than any input position, and the regression equations were forced to extrapolate ST estimates based on the input data. The errors resulting from this phenomenon highlight the importance of optimizing the set of regression calibration positions to encompass the entire desired test range of motion.

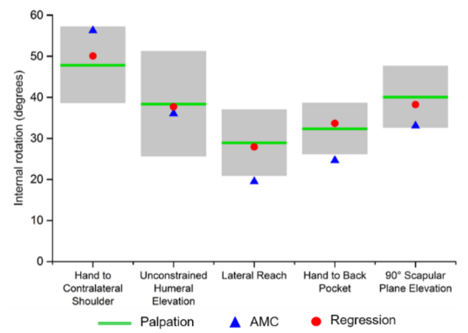

Figure 2: Mean ST internal rotation angles across all positions. Grey bars represent ± one standard deviation of palpated angles

While the double calibration mitigated AMC errors along the upward rotation axis, the device still significantly underestimated ST internal rotation. Abnormal motion of the scapula about this axis, i.e. scapular winging, can indicate scapular dysfunction [1], and accurate measurement of this motion is essential for diagnostic evaluation and outcome assessments. The limitations of the AMC in capturing this motion prohibit its use in populations where this motion is important. In conclusion, we recommend the use of the regression approach, as it is devoid of systematic errors around any axis and yielded smaller RMS errors for all positions.

REFERENCES

1. Ludewig, P. et al. (2009) Journal of Orthopedic Sport Physical Therapy, 39(2):90-104.
2. Karduna, A. et al. (2001) Journal of Biomechanical Engineering, 123(2):184-190.
3. Lempereur, M. et al. (2012) Journal of Biomechanics, 45(11):2028-2034.
4. Nicholson, K. (2015) University of Delaware, ProQuest Dissertations Publishing:10014746.
5. Brochard, S. et al. (2011) Journal of Biomechanics, 44(4):751-754.

DISCLOSURE STATEMENT

The authors have no conflicts of interest to disclose.

E-Posters

E-Poster Slide Presentations Available HERE at Open Conference GCMAS 2017: (https://openconf.org/GCMAS2017)

	TITLE	AUTHORS
1	Accuracy of a Low Cost, Portable Digital Video Analysis System to Advance Understanding of Global Spinal Motion	Anthony Petrizzo, Justin Paul, Thomas Errico, Ellen Godwin, Virginie Lafage
2	Changes in Gait Kinematics and Kinetics Following an Anterior Tibialis Transfer in Children with Recurrent Clubfoot	Jessica Lloyd, Pierz Kristan, Phil Mack, Jeffrey Thomson, Sylvia Ounpuu
3	Comparing the Accuracy and Reliability of 2D Video-Based and 3D Instrumented Gait Analysis for Kinematic Measures During Treadmill Running	Kyle Nagle, Kayla Burnim, Ariel Kiyomi Daoud, Susan Kanai, Matthew Sremba, Allison Frickman, Jason Rhodes
4	Comparision of Walking Versus Running Foot Kinematics with a Segmented Foot Model	Mark McMulkin, Bruce MacWilliams
5	Comparison of 3 Casting Methods for Custom Foot Orthoses	Jinsup song, Kersti Choe, Howard Palamarchuk, James McGuire, James Furmato
6	Creating Realtime Feedback For A Custom Virtual Reality System	Christopher Wilson, Samuel Chesebrough, Andrew Merryweather, K. Bo Foreman
7	Deficits in Mobility and Strength After Tibiotalar Arthrodesis	Jennifer Nichols, Andrew Anderson
8	Development of a Gait Score for the Assessment of End Stage Ankle Osteoarthrosis and Outcome of Related Surgery	Swati Chopra, Xavier Crevoisier
9	Difference in the Kick Motion of Adolescent Soccer Players Between Presence and Absence Of Low Back Pain	Michio Tojima, Suguru Torii
10	Effects of Body Weight Support Exercise by Spider Therapy on the Walking of a Cerebral Palsy Patient	Satomi Tada, Yasuhiko Hatanaka, Kouichi Saito, Kazuki Yamaguchi
11	Effects of Varying Upper Limb Usage and Resistance Settings on Muscle Activity During Elliptical Machine Exercise	Peter Quesada, Amanda Ritter
12	Functional Outcome Comparison Between 6 And 12 Month Follow-Up from ACL Reconstruction in Pediatrics	Vincent Yaccarino, Michael Tednes, Xiu Cheng Liu
13	Gait and Physical Activity of Urban-Living Individuals with Osteogenesis Imperfecta in Southeastern Asia	Nikhil Kurapati, Rebecca Boerigter, Carlo Sumpaico, Joycie Abiera, Melanie Alcausin, Peter Adamczyk, Peter A. Smith, Gerald F Harris
14	Immediate Effects and Aftereffects of Using a Gait Assist Suit Device When Walking Without the Device	Koichi Saito, Kazuki Yamaguti, Takaaki Nakamata, Tomomi Tada, Yasuhiko Hatanaka
15	Kinematic Differences Between Plug-In-Gait and a 6-Degree-of-Freedom Model During Walking and Cutting	Kyle Chadwick, Nicole Mueske, Tishya Wren

E-Posters (continued)

	TITLE	AUTHORS
16	Longitudinal Instrumented Gait Analysis on the Use of a Carbon Fiber AFO for a Child with Cerebral Palsy	Christina Bickley , Judi Linton, Douglas Barnes
17	*Muscle Activation Pattern during Gait and Stair Activities following Total Hip Arthroplasty with a Direct Anterior Approach: A Comprehensive Case Study*	*Swati Chopra, Michael Taunton , Kenton Kaufman*
18	*Recovery after Single Event Multi-Level Lower Extremity Surgery: Pilot Study*	*Ann Flanagan, Haluk Altiok, Nikhil Kurapati, Nancy Scullion, Mary Peer, Sahar Hassani*
19	*Relationship of Anthropometrics and Throwing Biomechanics in Youth Baseball Pitchers*	*Michael Apolinario, Shayne Fehr, Derek Damrow, Adam Hadro, Xue-Cheng Liu, Sergey Tarima*
20	*Reliability Assessment of Gait Kinematics in Pediatrics: A Single Case Study*	*Zahra Safaeepour, Mark Geil*
21	*The Problem of Multiple Comparisons Between Groups of Time Dependent Data*	*Timothy Niiler*
22	*Upper Extremity Biomechanical Analysis to Assess Thoracic and Pelvic Compensatory Strategies in Brachial Plexus Palsy Following Surgery*	*Micah Garcia, Alyssa Schnorenberg, Allison Algier, Emily Louden, Kevin Little, Lianna Hawi, Brooke Slavens, Jason Long*
23	*Use of Motion Analysis and SmartWheel to Assist in Wheelchair Prescription: A Case Study*	*Ann Flanagan, Haluk Altiok, Adam Graf, Nancy Scullion, Sahar Hassani*

ACCURACY OF A LOW COST, PORTABLE DIGITAL VIDEO ANALYSIS SYSTEM TO ADVANCE UNDERSTANDING OF GLOBAL SPINAL MOTION.

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INTRODUCTION
Dynamic measures of gait and motion performance have the potential to help surgeons determine what operative and radiologic parameters optimize human motion after surgery. For a more complete understanding of the effects of fusion on overall motion, studies of performance before and after surgery are required. The current gold-standard 3D motion capture can involve expensive, stationary equipment limiting access. This study sought to investigate the potential for an alternative low-cost, portable, 2D automated digital video motion analysis tool (Dartfish).

CLINICAL SIGNIFICANCE: The ability of a low-cost, portable motion analysis tool to accurately reproduce marker tracking in AIS patients during gait was examined. Simultaneously acquired video recordings analyzed with marker-tracking software (Dartfish) were compared with the gold standard 3D motion capturing system (Vicon) during the complex motion of gait.

METHODS: Each of 15 subjects with a diagnosis of AIS performed multiple gait analysis with simultaneous acquisition of two- and three-dimensional data as part of a cross-sectional clinical biomechanics study. Dartfish and Vicon generated data sets including the y (longitudinal) and z (vertical) marker position coordinates over an approximately 1 m path for 1.0 seconds during gait. The video generated positions were averaged for distance from the gold standard 3D position and the array of positions were compared for correlation and paired statistical t-test. Outcomes measured was the correlation and accuracy of spatiotemporal marker tracking

RESULT: Significant positive pairwise correlations ($p < 0.05$) were obtained when comparing 2D and 3D positions for 4 extremity markers in all 15 patients for a total of 60 tracked markers. For all markers, the overall average distance between Dartfish and Vicon generated positions was 5 +/- 3.2 cm. Accuracy improved for distal ankle and wrist markers than for proximal elbow and knee. Separately, the differences in the horizontal and vertical planes were 5 +/- 4.1 cm and 1.4 +/- 1.5 cm, respectively.

DISCUSSION: In the horizontal and vertical planes, 2D analysis accurately approximated the trajectory of upper and lower limbs during gait in 15 subjects with operative AIS. The horizontal component of the position contributed most to the error, which likely represents parallax at the beginning and end of the recorded gait. If so, accuracy should improve with stationary movement. Limited motion analysis data exist in spine surgery and are difficult to compare across studies. Taken together, the accuracy, affordability, portability and ease of use may offer a means toward higher-powered future studies of spine and pelvic range of motion.

CHANGES IN GAIT KINEMATICS AND KINETICS FOLLOWING AN ANTERIOR TIBIALIS TRANSFER IN CHILDREN WITH RECURRENT CLUBFOOT

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INTRODUCTION: There is little consensus on treatment strategies in children with recurrent clubfoot. Research evaluating surgical outcomes in patients with clubfoot is an important first step to achieving this goal; however, it is primarily focused on comparing post-operative data to a reference database. As a result, there is no understanding of pre-treatment level of deformity. Since the severity of clubfoot varies between patients, it is relevant to evaluate treatment outcomes in reference to pre-operative impairment which will ultimately lead to the establishment of evidence based treatment indications. Therefore, the purpose of this study is to better understand the outcomes of surgical treatment to correct recurrent clubfoot, specifically an anterior tibialis transfer (ATT), by comparing pre-operative and post-operative motion analysis assessments.

CLINICAL SIGNIFICANCE: Objective pre and post assessment of gait function in patients with clubfoot following an ATT using comprehensive motion analysis techniques will provide insight into the outcomes of this surgical intervention and ultimately put us closer to establishing evidence based treatment indications.

METHODS: Patients with clubfoot who underwent an ATT and had both a pre-operative (Evaluation 1) and post-operative (Evaluation 2) gait analysis were identified. For each evaluation, history, clinical exam, and motion data during gait were collected. Motion data were collected using a Vicon MX system (VICON, Los Angeles, CA) and ground reaction forces were collected using five force plates (AMTI, Watertown, MA) following standard procedures [1]. The heel bisector, a measure of forefoot adductus, was recorded as the location of a line bisecting the heel with respect to the metatarsals (ex. heel bisector of 3.5 passes between metatarsals #3 and #4). Manual muscle strength measures for the ankle were assessed using a 0 to 5 scale [2]. A paired t-test with an alpha of 0.05 was used to determine if there were significant differences in data between evaluations.

RESULTS: A total of 9 patients who have undergone an ATT on 10 affected clubfeet (mean age 5.3 ± 2.0 years at Evaluation 1 and 7.1 ± 2.1 years at Evaluation 2) were included. The surgical intervention of the affected clubfeet at the time of the ATT included the following: ATT only (2/10); ATT with other joint sparing procedures (3/10), ATT with posterior release (2/10), and ATT with posterior medial release (3/10). The

Table 1. Clinical Exam Data at Evaluation 1 and Evaluation 2 for each Case of Clubfoot

Case # (*same patient)	Dorsiflexion ROM with the Knee Extended (°)			Heel Bisector (Metatarsal Number)			Dorsiflexion Strength with the Knee Extended		
	Eval 1	Eval 2	Δ	Eval 1	Eval 2	Δ	Eval 1	Eval 2	Δ
1	-20	-5	15	4.0	4.0	0.0	4	4	0
2	5	0	-5	3.5	3.0	-0.5	5	5	0
3*	0	0	0	5.0	5.0	0.0	4	5	1
4*	5	0	-5	5.0	4.5	-0.5	4	5	1
5	0	5	5	5.5	5.0	-0.5	n/a	5	n/a
6	0	0	0	5.0	3.5	-1.5	5	4	-1
7	-5	5	10	4.5	3.5	-1.0	n/a	5	n/a
8	5	15	10	4.5	4.0	-0.5	n/a	5	n/a
9	0	0	0	3.5	3.0	-0.5	5	5	0
10	-5	5	10	5.5	5.0	-0.5	n/a	5	n/a
Mean/ Median	**-2**	**3**	**4**	**4.6**	**4.1**	**-0.6**	**5**	**5**	**0**
SD/ Mode	**7**	**5**	**7**	**0.7**	**0.8**	**0.4**	**4**	**5**	**0**
t-test E1 v E2	**p=0.1039**			**p=0.0032**			**p=0.6109**		

average time between surgery and Evaluation 2 was 1.5 ± 0.7 years. Significant differences ($p<0.05$) between Evaluation 1 and Evaluation 2 were noted in the heel bisector ($p=0.0032$) (Table 1), peak dorsiflexion moment during loading response ($p=0.0212$), peak dorsiflexion during stance ($p=0.0056$), and peak ankle plantar flexion moment during stance ($p= 0.0361$) (Table 2).

Table 2. Kinematic and Kinetic Data at Evaluation 1 and Evaluation 2 for each Case of Clubfoot

Case # (*same patient)	Mean Foot Progression during Stance (°, +int/-ext)			Peak Dorsiflexion during the Middle Third of Swing (°, +dorsi/-plantar)			Ankle Angle at 98% of the Gait Cycle (°, +dorsi/-plantar)			Peak Dorsiflexion Moment during Loading Response (Nm/kg, +plantar/ -dorsi)			Peak Dorsiflexion during Stance (°, +dorsi/-plantar)			Peak Ankle Plantar Flexion Moment during Stance (Nm/kg, +plantar/ -dorsi)		
	Eval 1	Eval 2	Δ	Eval 1	Eval 2	Δ	Eval 1	Eval 2	Δ	Eval 1	Eval 2	Δ	Eval 1	Eval 2	Δ	Eval 1	Eval 2	Δ
1	27	-5	-33	-13	-13	0	-10	-14	-4	-0.01	-0.01	0.00	-4	3	7	0.76	1.06	0.30
2	3	6	3	-1	2	3	1	2	1	-0.02	-0.11	-0.09	8	18	10	0.00	0.66	0.66
3*	6	2	-3	-1	-3	-2	-2	-1	2	n/a	n/a	n/a	12	13	1	n/a	n/a	n/a
4*	8	9	1	2	1	-1	4	1	-2	n/a	n/a	n/a	11	13	3	n/a	n/a	n/a
5	7	7	0	-4	2	6	-12	3	15	n/a	-0.04	n/a	7	14	6	n/a	0.66	n/a
6	-3	-22	-19	1	-8	-9	2	-13	-15	-0.01	-0.02	-0.02	14	16	2	0.69	0.54	-0.15
7	-6	7	13	-10	-4	5	-3	-5	-2	-0.01	-0.07	-0.06	8	18	10	0.78	1.15	0.38
8	10	6	-4	3	1	-2	0	1	1	n/a	-0.10	n/a	18	16	-2	n/a	0.86	n/a
9	13	7	-6	-4	-5	-1	-17	-1	16	-0.01	-0.10	-0.09	7	11	3	0.56	1.05	0.49
10	36	7	-29	5	-2	-6	3	2	-1	-0.01	-0.14	-0.13	14	17	4	0.59	0.83	0.24
Mean	**10**	**2**	**-8**	**-2**	**-3**	**-1**	**-4**	**-2**	**1**	**-0.01**	**-0.07**	**-0.07**	**9**	**14**	**4**	**0.56**	**0.85**	**0.32**
SD	**13**	**9**	**15**	**5**	**5**	**5**	**7**	**6**	**9**	**0.00**	**0.05**	**0.05**	**6**	**4**	**4**	**0.29**	**0.22**	**0.28**
t-test E1 v E2	**p=0.1280**			**p=0.6570**			**p=0.7165**			**p=0.0212**			**p=0.0056**			**p=0.0361**		

DISCUSSION: The significant improvement in the heel bisector following surgery indicates an improvement in metatarsus adductus as a result of this combination of procedures. The impact of the ATT can be more specifically assessed by evaluating changes in dorsiflexion strength and ankle function during swing phase. Dorsiflexion strength remained the same in 5/6 sides with pre and post-op data and was full (score of 5/5) following surgery in the remaining four sides without pre-op data, which is a counter point to the commonly held belief that surgical transfer results in a reduction of one grade in muscle strength. The lack of change in ankle angle in mid swing and at 98% of the gait cycle, two additional measures of functional dorsiflexor strength, also suggests that anterior tibialis strength was not compromised due to the ATT. The greater incidence of a dorsiflexor moment during loading response following surgery indicates an increased incidence of heel initial contact, a pre-requisite of typical gait. The improved peak plantar flexor moment may be a functional outcome of the reduced adductus following surgery, which allows for a better lever arm in the direction of progression.

This preliminary data suggests that the ATT does not result in increased weakness nor the associated gait issues of increased equinus in swing and at initial contact. Further research is needed to determine if the improvements in adductus and ankle moments hold true for a larger sample size and were due to the ATT alone.

REFERENCES:

1. Davis, R. et al. (2000) Analysis Biomedical Engineering Handbook, pp. 1-9.
2. Kendall F. et al. (1983) Muscle Testing and Function, pp. 23.

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COMPARING THE ACCURACY AND RELIABILITY OF 2D VIDEO-BASED AND 3D INSTRUMENTED GAIT ANALYSIS FOR KINEMATIC MEASURES DURING TREADMILL RUNNING

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INTRODUCTION

This pilot study aimed to demonstrate the accuracy of two dimensional (2D) video analysis compared to three dimensional (3D) analysis of sagittal and coronal plane kinematics during treadmill running.

CLINICAL SIGNIFICANCE

2D remains a useful tool for qualitative analysis of running mechanics. For clinical quantification and for research, 3D remains more accurate and should be used.

METHODS

Subjects, ages 21-35, 9 female, 8 male, were instrumented for 2D and 3D motion capture. 2D and 3D video data were simultaneously collected. Five 2D raters reviewed gait cycles from 3 trials per subject to measure sagittal trunk, hip, knee, and ankle angles at initial contact and mid-stance, as well as coronal view mid-stance lateral pelvic tilt, hip, and knee angles. These same measurements were obtained from 3D motion capture using MatLab. 2D and 3D measurements were compared using Bland-Altman Analysis of Agreement and a mixed effects model. 2D and 3D agreed if the limits of agreement (LOA) outside the mean difference between 2D and 3D (the bias) were less than the predetermined clinically significant limit (5 degrees for all angles).

RESULTS

2226 total angle data points were obtained (2030 in 2D, 196 in 3D). In the sagittal plane, during initial contact, 2D measurements of only trunk (bias=1.86, 1.93, 2.18; LOA= -3.54-4.02, -4.92-3.61, -4.43-2.87) and hip (bias= -1.56, -1.44, -0.94; LOA= -4.86-1.74, -4.91-2.03, -5.51, 3.63) agreed with 3D measurements. During mid-stance, trunk angles agreed in two trials (bias= 0.35, 0.65, 0.92; LOA = -4.37-5.07, -4.63-5.94, -3.78-5.62) while hip, knee, and ankle angles did not agree. In mid-stance coronal views, hip angles agreed (bias= -0.25, -0.79, -0.71; LOA= -4.54-4.03, -4.09-2.50, -5.27-3.84) whereas pelvis tilt angles only agreed for one trial. Anterior mid-stance 2D knee angles did not agree with 3D.

DISCUSSION

The data suggests that limited angle measurements agree between 2D and 3D analysis during treadmill running.

REFERENCES

1. van der Worp MP, ten Haaf DSM, van Cingel R, de Wijer A, Nijhuis-van der Sanden MWG, Staal JB. Injuries in Runners; A Systematic Review on Risk Factors and Sex Differences. PloS one. 2015;10(2):1-18.
2. Renstrom AF. Mechanism, diagnosis, and treatment of running injuries. Instructional course lectures. 1993;42:225-234.
3. Louw M, Deary C. The biomechanical variables involved in the aetiology of iliotibial band syndrome in distance runners – A systematic review of the literature. Physical Therapy in Sport. 2014;15(1):64-75.
4. van Gent RN, Siem D, van Middelkoop M, van Os AG, Bierma-Zeinstra SM, Koes BW. Incidence and determinants of lower extremity running injuries in long distance runners: a systematic review. British journal of sports medicine. 2007;41(8):469-480; discussion 480.
5. Taunton JE, Ryan MB, Clement DB, McKenzie DC, Lloyd-Smith DR, Zumbo BD. A retrospective case-control analysis of 2002 running injuries. British journal of sports medicine. 2002;36(2):95-101.
6. Tiberio D. The effect of excessive subtalar joint pronation on patellofemoral mechanics: a theoretical model. The Journal of orthopaedic and sports physical therapy. 1987;9(4):160-165.
7. DeLeo AT, Dierks TA, Ferber R, Davis IS. Lower extremity joint coupling during running: a current update. Clinical Biomechanics. 2004;19(10):983-991.
8. Powers CM. The influence of abnormal hip mechanics on knee injury: a biomechanical perspective. The Journal of orthopaedic and sports physical therapy. 2010;40(2):42-51.
9. Powers CM. The influence of altered lower-extremity kinematics on patellofemoral joint dysfunction: a theoretical perspective. The Journal of orthopaedic and sports physical therapy. 2003;33(11):639-646.

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COMPARISON OF WALKING VERSUS RUNNING FOOT KINEMATICS WITH A SEGMENTED FOOT MODEL

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INTRODUCTION

Quantitative three dimensional kinematic and kinetic analysis for typical walking, running, and sprinting have been provided for the pelvis, hip, knee and ankle [1]. A three dimensional motion analysis comparing running in addition to walking for children with cerebral palsy and healthy control children has also been reported [2]. However, a segmented foot model analysis for running has not previously been reported. Therefore, the purpose of this study is to compare the three dimensional kinematic motions of the tibia, hindfoot, forefoot, and hallux for running versus walking in typically developing children.

CLINICAL SIGNIFICANCE

Kinematic analysis of running can be an important addition to the standard walking kinematic and kinetic testing procedures currently completed for decision making. Understanding any differences in segmented foot model kinematics between walking and running of typically developing children has the potential to assist treatment decisions for pathological foot issues.

METHODS

Institutional review board approval was received to conduct a larger study to establish typical movement data (physical exam, pedobarography, kinematics, kinetics, EMG, energy consumption testing) for clinical testing for a single motion lab (Spokane Movement Analysis Center). As part of this study, a segmented foot model [3] was employed to assess foot kinematics for walking and running activities. The standard Plug-in-Gait marker set along with additional foot model markers were placed on the subjects. Data was captured using a 12 camera Vantage (V16) Vicon camera system and 4 AMTI force plates. Standard lower extremity kinematics and kinetics were collected. Only segmented foot kinematics are reported here. These include tri-planar motions of the hindfoot relative to the tibia (ankle complex) and forefoot relative to hindfoot (midfoot) and 2D motions of the hallux relative to forefoot.

Subjects were asked to walk at a self-selected speed and data was collected until six clean forceplate strikes were obtained for each foot. Subjects were then asked to run across the lab (length of 40 feet) and data was collected until a single clean force plate strike for each foot was collected. Data for one randomly selected walking trial was used for comparison to the running trial. Paired t-tests were used to compare walking versus running for variables of min, max and range across the gait cycle of the three planes of motion for ankle complex and midfoot and two planes of motion for the hallux. Significance was set at the level of 0.05.

RESULTS

17 subjects participated in this study (age range 5 to 21 years, mean 13.2). Mean self-selected walking speed was 1.24 m/sec and mean running speed was 3.20 m/sec, the same as those

previously reported [1]. Mean values for segmented foot motions were significantly different between running and walking (Table 1). Most notably running led to increased ankle complex eversion, ankle complex dorsiflexion, ankle complex external rotation, and midfoot flexion. Walking led to greater midfoot inversion, midfoot adduction, and hallux extension.

Table 1. Summary of segmented foot model kinematic variables between walking and running. Bold p-values considered significant differences.

	Walking			**Running**			**p value**		
Variable (all units Degrees)	Min	Max	Range	Min	Max	Range	Min	Max	Range
Ankle Complex Motions									
Inversion (+)/Eversion (-)	-4.9	8.6	13.5	-7.7	7.6	15.0	**<0.001**	0.21	**0.045**
DF (+)/PF (-)	-0.3	26.2	26.5	-1.3	33.8	35.1	0.29	**<0.001**	**<0.001**
Int (+)/Ext (-)Rotation	-2.8	8.4	11.2	-6.1	7.0	13.1	**<0.001**	0.06	**<0.001**
Midfoot Motions									
Inversion (+)/Eversion (-)	-7.4	5.4	12.8	-7.0	3.8	10.8	0.44	**0.01**	**0.03**
Flexion (+)/Extension (-)	-48.7	-27.0	21.8	-47.8	-22.7	25.1	0.18	**<0.001**	**<0.001**
Adduction (+)/Abduction (-)	2.2	15.7	13.5	1.5	11.5	10.0	**0.00**	**<0.001**	**<0.001**
Hallux Motions									
Extension (+)	6.5	73.9	67.3	5.0	64.3	59.2	**0.00**	**<0.001**	**<0.001**
Varus (+)/Valgus (-)	-15.8	-1.5	14.3	-16.8	-3.0	13.8	0.11	**<0.001**	0.91

DISCUSSION

The purpose of this study was to assess segmented foot kinematics between walking and running with results generally indicating running led to increased ankle complex motions, greater midfoot flexion, and more limited midfoot inversion and adduction. Hallux extension was actually increased during walking. Higher forces/moments might largely explain differences in foot motions. Sagittal plane differences were 4-10°, with running increases in ankle complex dorsiflexion and midfoot flexion which could be explained by greater sagittal plane moments. However, running had lower hallux extension indicating perhaps an active mechanism to increase the overall moment arm to the end of the foot. Significant frontal plane differences were smaller (2-3°), with running leading to greater ankle complex eversion and limiting midfoot inversion. Finally, transverse plane differences were also smaller (1-4°), with running having greater ankle external rotation and more limited midfoot adduction and hallux varus. Smaller differences in frontal and transverse planes are likely related to overall smaller motion ranges. In conclusion, running did lead to several segmented foot motions that are significantly different that walking for typically developing subjects indicating potential clinical utility.

REFERENCES: **1**. Novacheck 1998, Gait Posture 7:77-95. **2**. Davids et al 1998, Dev Med Child Neurol 40:528-35. **3**. Saraswat et al 2012, Gait Posture 35:547–55.

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Comparison of 3 casting methods for custom foot orthoses

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INTRODUCTION

Custom foot orthoses (CFO) are commonly used to manage foot problems associated with aberrant foot biomechanics, including heel pain syndrome, stress fractures, and diabetic foot ulcers.[1-6] CFOs are traditionally made from a cast, a 3-dimensional foot shape captured in non-weight bearing condition using plaster of Paris. Recently, 3 dimensional laser scanners are being used to capture the foot shape. In addition, Vertical Foot Alignment System developed a novel method to directly fabricate CFO in weight bearing corrected alignment.[7] However, performances of these new casting techniques have not been objectively evaluated.

CLINICAL SIGNIFICANCE

CFO is commonly used to manage many foot pathologies. Biomechanical assay is needed to advance the casting, fabrication, and clinical utility of this common treatment modality.

METHODS

Healthy asymptomatic subjects, between ages 18-65 years with moderate pes planus, provided consent and participated in the study. Three pairs of CFOs were casted by designated experienced clinicians using 3 methods: plaster of Paris (P), a 3D laser scanner (Q), and weight bearing corrected molding technique (V). CFOs (P and Q) were fabricated in standard manner by the same orthotic laboratory. Shoe comfort rating[8] and dynamic in-shoe plantar pressure were measured (novel pedar-X, sampled at 100 Hz) during comfortable self-selected walking speed while wearing a standard shoe (New Balance, #574) and the same shoe with each of 3 pairs of CFO following 10-minutes of accommodation.

Descriptive statistics and normality testing were performed using SPSS software version 22 (IBM, Chicago, IL, USA). The lower limb was used as the unit of observation instead of the individual. A Generalized Linear Model with an identity link function was used to test difference across 4 shod conditions while accounting for potential dependence in bilateral data. The Wald Chi-square was calculated for each dependent variable with significance set at $p < 0.05$. Post hoc pairwise comparisons for all pairs were performed using the Generalized Chi Square test at $P < 0.05$.

RESULTS

Participants consisted of 24 subjects (9 female) with mean age of 25.4 ± 3.86 years old and BMI of 25.6 ± 3.79 kg/m^2. Resting calcaneal stance position (RCSP) showed significantly greater valgus hindfoot alignment in barefoot compared to barefoot standing over CFOs, see Table 1. Overall shoe comfort was significantly higher in V condition than other 3 shod conditions. Sneaker only (C) yielded the highest force time integral (FTI, Ns) under the metatarsal 1 (P=0.013) and the smallest FPI under medial arch (P=0.000). No significant difference is noted between P and Q conditions in any regions. CFO made from weight bearing corrected method (V) yielded a significantly lowered FTI under metatarsal 2-4, lateral arch, and heel regions than P and Q shod conditions.

Table 1: Summary of static hindfoot alignment, overall shoe comfort (%), and force time integral (FTI, in Ns) for different foot regions.

	Mean By Condition					
	C	P	Q	V	p-value	Note
RCSP (° valgus)	13.03	9.79	10.11	9.82	0.006	a,b,c
Shoe Comfort, %	70.5	64.9	65.6	78.4	0.041	c,e,f
FTI (Ns), Toe 1	23.9	27.1	26.2	24.6	0.106	
Metatarsal 1	23.0	20.1	19.3	20.2	0.013	a,b,c,e,f
Metatarsal 2	18.9	18.9	20.0	17.3	0.024	e,f
Metatarsal 3	16.6	18.3	19.3	16.4	0.007	a,b,e,f
Metatarsal 4	13.3	14.8	15.8	13.4	0.010	a,b,e,f
Metatarsal 5	10.5	11.0	11.7	10.5	0.198	
Arch, medial	13.8	22.5	21.8	19.3	0.000	a,b,c
Arch, lateral	35.5	33.6	34.2	28.0	0.006	c,e,f
Heel, medial	53.9	52.4	50.5	46.4	0.011	b,c,e,f
Heel, lateral	50.2	51.2	52.4	44.3	0.007	c,e,f

Note: Resting Calcaneal Stance Position (RCSP) in barefoot and 3 CFOs. Mean values of shoe only (C), CFO casted using Plaster of Paris (P), 3D laser scanner (Q), and weight bearing corrected (V) are shown. A significant difference in pairwise post hoc comparison between C and P (a), C and Q (b), C and V (c), P and Q (d), P and V (e), and Q and V (f) are denoted.

DISCUSSION

Various theories and techniques exist for casting, fabrication, and utilization of CFOs. Two novel casting methods were compared to the standard of care. No significant differences were noted between the traditional (P) and a 3D laser scanning (Q) based on shoe comfort rating and in-shoe plantar pressure assessment in 24 healthy subjects with moderate pes planus. Weight bearing neutrally aligned direct molded CFO (V) yielded significantly greater shoe comfort and reduced total load in many regions of foot. Additional studies are needed to refine the methods and clinical efficacy of CFO.

REFERENCES

1. Finestone, A., et al. (1999) Clin Orthop Relat Res, 360: 182-90.
2. House, C. et al. (2013) Mil Med, 2013. 178(6): p. 683-9.
3. Budiman-Mak, E., et al. (1995) J Clin Rheumatol, 1(6): p. 313-322.
4. Wrobel, J.S., et al. (2013) J Am Podiatr Med Assoc, 105(4): p. 281-94.
5. Barn, R., et al. (2014) Rheumatology (Oxford), 53(1): p. 123-30.
6. Ulbrecht, J.S., et al. (2014) Diabetes Care, 37(7): p. 1982-9.
7. Payne, C., et al. (2001) Australasian Journal of Podiatric Medicine, 35(3): p. 65-71.
8. Che, H. et al. (1994)Clin Biomech (Bristol, Avon), 9(6): p. 335-41.

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Performance Laboratories Inc. manufactured CFOs (P and Q) and Vertical Orthotics Pty Ltd provided supplies to fabricated CFO. Jinsup Song serves as a medical consultant for Vertical Orthotics.

Creating Realtime Feedback for a Custom Virtual Reality System

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INTRODUCTION

Virtual Reality (VR) has grown significantly over recent years as a tool for both research and rehabilitation. This is in part due to the interactive and challenging environments that are created that have fewer environmental risks as compared to the real world. To alter these VR environments, the amount of immersion and interaction the user encounters is changed. This can be accomplished by changing the visual graphics inside the virtual environment and/or by adding or changing hardware configurations. However, most of the components of the system are closely integrated and getting all of the systems to be synchronized and work together is often difficult. Therefore, our objective is to detail the methods we used to develop our Cave Automatic Virtual Environment (CAVE) system for both training and data collection.

CLINICAL SIGNIFICANCE

VR systems are becoming more affordable and their use as a rehabilitation tool more common. This is a convenient method of training since VR can bring the environment to the user instead of taking the user to the environment. In addition, the environment can be made more or less difficult with adjustment to the software making it more or less challenging to a single users and adaptable to different users. Therefore, it is important to develop the appropriate methodology to assure proper functioning of VR systems for training and data collection.

METHODS

The VR system used at the University of Utah is a Cave Automatic Virtual Environment (CAVE) system referred to as the Treadport. At the center of the Treadport is a large (10ft long by 8ft wide) treadmill with custom control software and a high-resolution encoder which is accurate to +/- 0.05 m/s during normal loading. The control software takes into account the belt-deck friction and user weight to ensure proper belt torque and constant belt speed. The custom control software allows for the user to walk freely on the treadmill, speeding up and slowing down as needed.[1] The virtual world is projected on three 8ft-square rear-projection screens and onto the belt surface using Optoma EH505 projectors. These projectors are capable of producing 3D projections with a 3D resolution of 960x1200 pixels per screen at 60 Hz per eye.

Users interact with the virtual world using Vicon's real-time motion tracking capabilities (Vicon, Oxford, UK). Using Vicon's real-time software development kit (Realtime SDK) and University of North Carolina at Chapel Hill's open-source Virtual Reality Private Network (VRPN) library, a C++ program was developed to broadcast the marker and segment locations and orientations over a TCP network connection from the motion capture computer to the computer running the virtual reality simulation.

The virtual reality simulation was created in the Unity game engine (Unity Technologies San Francisco, CA). The Unity engine allows for rapid development of environments and assets to create immersive virtual environments. Using C#, a script was developed to read and parse the VRPN data from the Vicon system and assign the marker and segment positions within the virtual world. This allowed the segments of the user to interact with the objects created using the Unity game engine. Unity comes with a physics engine to handle in-game collisions of virtual objects. Using this physics engine, and appropriate mass estimations for human feet and other appendages, realistic interactions between the user and the virtual world were simulated in real time.

In order to properly couple the movement of the subject with the virtual objects, the origin and coordinate system of both the physical and virtual world needed to by synchronized. The coordinate system of the physical world was defined by capturing the position of a calibration frame using the motion capture system. The coordinate system of the virtual world was then aligned to the real-world with a more complicated process that verified the relative positions of a physical object and virtual object. This was accomplished by placing a physical box, that was instrumented with reflective markers, in the virtual world and then creating an identical box in the virtual world and shifting the coordinate systems until they were aligned. A script was developed to record the position and time-stamp (Windows system time: year-month-day-hour:minute:second:millisecond format) of the virtual box position within this reference frame and at each update of the physics engine in a format conducive to importing the box trajectories into Visual 3D. In order to synchronize the timelines between the Vicon system (constant rate) and the virtual-world recording system (virtual rate) an additional event needed to be detected. This event was the toggling of the LED lights on a calibration wand. When the wand was activated, a flag was placed simultaneously in the VR recording file and the Vicon motion file. The use of this flag allowed for synchronization between using Visual 3D v5.01.15 (C-motion, Inc., Germantown, MD). Once that was complete, the ability to analyze a participant in the virtual environment in relation to virtual objects was possible.

SUMMARY

The use of Virtual Reality in rehabilitation and research settings is becoming more common then ever before. As this happens, the ability to appropriately synchronize data from the various systems being used for that VR system is imperative to ensure accuracy during interaction within the virtual environment and accuracy of the data for post processing.

REFERENCES

1. Hejrati B, Crandall KL, Hollerbach JM, Abbott JJ. Kinesthetic Force Feedback and Belt Control for the Treadport Locomotion Interface. IEEE transactions on haptics. Apr-Jun 2015;8(2):176-187.

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DEFICITS IN MOBILITY AND STRENGTH AFTER TIBIOTALAR ARTHRODESIS

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INTRODUCTION

Tibiotalar arthrodesis, which involves fusing the tibia and talus, is the most common treatment for ankle osteoarthritis. Clinical studies generally report patient satisfaction following surgery [1]. However, these positive outcomes are due to successful pain relief, not restored function. Notably, following tibiotalar arthrodesis patients exhibit inefficient gait, which is primarily contributed to restrictions in ankle range of motion [2]. Other factors such as deficits in ankle strength, which may also contribute to altered gait, are not fully understood.

The objective of this study was to evaluate the extent to which ankle range of motion and ankle strength were altered following tibiotalar arthrodesis. Given that the tibiotalar joint is primarily responsible for dorsiflexion-plantarflexion motion, we hypothesized that compared to the contralateral limb, the surgically-altered limb would exhibit larger deficits in range of motion and strength for dorsiflexion-plantarflexion than for inversion-eversion.

CLINICAL SIGNIFICANCE

This study will inform clinical innovations, such as targeted rehabilitation programs, to restore mobility and strength to patients following tibiotalar arthrodesis.

METHODS

Seven subjects with isolated, unilateral tibiotalar arthrodesis (3 female, 55.3 ± 7.5 years old, BMI 30.4 ± 3.4) (mean ± SD) participated in this IRB approved study. All subjects were greater than 12-months post-operative with radiographic evidence of boney fusion.

Active range of motion and voluntary isometric ankle joint torque were measured in both limbs of each subject for four directions of motion: inversion, eversion, dorsiflexion, and plantarflexion. All measurements were acquired using a Human Norm Isokinetic Extremity System (CSMI, Stoughton, MA). During testing, subjects were prone with their foot tightly secured to a load cell via a metal plate with their knee and hip in 60º flexion. For the isometric torque testing, subjects exerted maximal force in each direction with their ankle in a neutral posture. Verbal encouragement was provided. Three, three-second trials were recorded for each direction. The maximum torque achieved across all trials was used for analysis A trial was excluded from analysis if the maximum torque recorded in that trial varied from the average torque recorded over all trials by more than 15 percent. All trials excluded using this criterion exhibited a maximum torque value that was substantially larger in magnitude than the other two recorded trials, suggesting that subjects utilized their hip and/or knee during the excluded trials.

For both the range of motion and maximum isometric torque measurements, paired two-tailed t-tests (significance level, $p < 0.05$) determined if significant differences existed between the surgically-altered limb and the contralateral limb on which surgery was not performed.

RESULTS

The range of motion in the surgically-altered limb was significantly ($p < 0.01$) smaller than that of the contralateral limb for dorsiflexion-plantarflexion and inversion-eversion (Table 1).

Importantly, dorsiflexion-plantarflexion demonstrated a significantly larger reduction ($p = 0.004$) in range of motion than inversion-eversion. Across subjects, mean (± SD) dorsiflexion-plantarflexion range of motion in the surgically-altered limb was 35.0 ± 8.5% of that in the contralateral limb. In contrast, mean (± SD) inversion-eversion range of motion in the surgically-altered limb was 54.2 ± 9.6% of that in the contralateral limb.

The surgically-altered limb was significantly weaker ($p < 0.01$) than the contralateral limb for dorsiflexion, inversion, and eversion (Table 1). The largest strength deficit was found in dorsiflexion, where the mean (± SD) isometric torque produced by surgically-altered limb was 51.5 ± 15.7% of that produced by the contralateral limb. Interestingly, the large standard deviations for the maximum isometric torque indicated that not all subjects displayed a substantial decrease in strength following tibiotalar arthrodesis.

Table 1. Measurements in Limb with Fused Ankle as Percentage of Contralateral Limb

Subject	Range of Motion		Maximum Isometric Torque			
	DF/PF	Inv/Ev	DF	PF	Inv	Ev
1	25.4	39.1	41.9	39.9	76.3	51.1
2	35.3	71.0	51.2	78.5	96.9	91.7
3	31.0	55.1	54.1	49.9	33.0	55.4
4	29.8	56.4	47.5	31.3	96.7	88.0
5	30.2	52.0	39.4	89.0	42.2	70.1
6	45.7	56.9	41.6	62.2	74.3	88.5
7	47.5	49.1	84.9	85.1	42.0	55.8
Mean (± SD.)	**35.0 ± 8.5**	**54.2 ± 9.6**	**51.5 ± 15.7**	**62.3 ± 22.8**	**65.9 ± 26.8**	**71.5 ± 17.8**

DISCUSSION

The range of motion deficits observed in this study are similar to those reported in the literature [3]. To our knowledge, isometric torque-generating capacity following tibiotalar arthrodesis has not been previously reported. Our torque results suggest that for most patients the torque-generating capacities of the dorsiflexor and invertor muscles are affected to a larger extent than those of the plantarflexor and extensor muscles. Limited dorsiflexion strength could hinder toe clearance during the transition from stance to swing and negatively influence the forward progression of the foot during stance. This interpretation is supported by studies reporting decreased stride length and cadence following tibiotalar arthrodesis [2]. Future work in our lab will utilize imaging data from the subjects examined in this study to determine whether changes in muscle volume contribute to the observed deficits in mobility and strength.

REFERENCES

1. Ajis, A. et al. (2013) Foot & Ankle International, 34(5):657-665.
2. Thomas, R. et al. (2006) Journal of Bone & Joint Surgery, 88(3):526-535.
3. Sealey, R.J. et al. (2009) Foot & Ankle International, 30(3):187-196.

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DISCLOSURE STATEMENT

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Development of a Gait Score for the Assessment of End-stage Ankle Osteoarthrosis and Outcome of Related Surgery

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INTRODUCTION

Subjective patient reported outcome scores (PROS) have been known to overestimate the outcome of patients following orthopaedic surgeries [1, 2]. Objective gait assessment methods, on the other hand, are found to be more reliable dat but have limitations for clinical use due to cost, complexity, time and space requirements. This study aimed to simplify gait assessment by pinpointing the most clinically relevant gait parameters and developing a score out of the reduced parameter set. The study also utilized an ambulatory gait assessment system which is cheaper, portable and easy to use.

CLINICAL SIGNIFICANCE

- Simplification of gait assessment for clinical use
- The developed score could be used as an objective tool of assessment in rehabilitation
- Ambulatory gait assessment provides freedom to test patient in an open environment

METHODS

89 participants were assessed using ambulatory gait assessment method, including 24 controls, 15 with end-stage ankle osteoarthritis (AOA), 15 with ankle arthrodesis (AA), 20 with total ankle replacement (TAR) and 15 with tibiotalocalcaneal arthrodesis (TTCA). AOFAS (American Orthopaedic Foot and Ankle Society) and FAAM (Foot and Ankle Ability Measure) questionnaires were filled out by the patients. Altogether, 48 gait parameters were assessed including spatiothemporal, kinematic and plantar pressure parameters. All parameters showing a significant difference ($p<0.05$) from the controls were used in a principal component analysis (PCA). Hotelling's T^2-statistic, a goodness of fit test, was used to determine those most statistically significant data, reducing the parameter set without losing any clinically relevant information. Lastly, using the reduced parameter set, a gait score was developed using Tukey's honest significance method. Based on this method, each of the selected parameters is then scored according to their difference with the controls. The final result is then normalized to provide a comprehensive gait score out of 100.

RESULTS

17 parameters were identified that provided a notably very high correlation to the full set of 48 ($r=0.91$, $p<0.001$). The reduced parameter set included cadence, speed, peak swing speed, toe-off pitch angle, sagittal plane motion at toe-forefoot, forefoot-hindfoot and forefoot-shank

intersegments, coronal plane motion at forefoot-shank intersegment, total contact at hindfoot lateral, midfoot lateral, forefoot central, hindfoot lateral, forefoot central and first toe, maximum force at hindfoot medial, second and first toe and lastly, maximum pressure at hindfoot medial and first toe, with total contact at hindfoot lateral and forefoot central. The clinical and gait scores for the 4 case groups are given in (Table 1). Little to no correlation was found between final gait score and the clinical scores from all 4 case groups (Table 2).

Table 1: Gait, FAAM and AOFAs scores for case groups. Median (IQR)

Groups	GAIT Score	FAAM	AOFAS
AOA	64 (17)	59.5 (25.5)	55.5 (27.75)
AA	67 (10)	67.5 (25.6)	66 (8)**
TAR	74 (4.5)*	83.8 (18.8)**	83.5 (13.25)**
TTCA	71 (12)	70 (25.5)	63 (11.5)*

* $p<0.05$, ** $p<0.01$ compared to the AOA group

Table 2: Correlation (r) between the scores for each of the four case groups

Groups	Gait score vs FAAM	Gait score vs AOFAS	FAAM vs AOFAS
AOA	0.07 (p=0.8)	0.28 (p=0.5)	0.40 (p=0.3)
AA	0.46 (p=0.15)	0.05 (p=0.9)	0.14 (p=0.7)
TAR	0.08 (p=0.8)	0.36 (p=0.2)	0.79 (p<0.001)
TTCA	0.16 (p=0.6)	0.61 (p=0.05)	0.43 (p=0.2)

DISCUSSION

A predictive and robust model for quantifying patient ankle function was developed. PCA was used to identify 17 parameters that provide clinically relevant information. Furthermore, the rationale for each of the parameters can be justified based on the outcome of previous gait studies. For example, speed and cadence have been considered important in assessing health status of patients with chronic illness and peak swing speed and toe-off pitch angle play an important role in maintaining both speed and angular momentum during walking. Moreover, the sagittal plane performs the majority of motion for propelling the body forward with each section of the foot being equally represented within this plane. The authors conclude that that the developed gait score could indeed simplify objective gait assessment in a clinical setting without misrepresenting or foregoing relevant patient data. Further research, i.e. clinical trials, is however required to test the reliability and the robustness of the developed gait score.

REFERENCES

1. Noble PC et al. Clinical Orthopaedic Related Research, 471(11):3437-45.
2. Parker J et al. (2003) Foot and Ankle International, 24(1):56-60.

DISCLOSURE STATEMENT

No conflicts of interest to disclose.

DIFFERENCE IN THE KICK MOTION OF ADOLESCENT SOCCER PLAYERS BETWEEN PRESENCE AND ABSENCE OF LOW BACK PAIN

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INTRODUCTION

Many adolescent athletes experience low back pain (LBP), which is the third most common soccer-related disorder. The odds ratio for soccer-related LBP is 1.8 [1]. Adolescent soccer players who maturated late have more soccer-related disorders than players who maturated early [2].

During the kick motion, adolescent soccer players need to place their support leg near the ball to rapidly shoot the ball. However, there are no reports on the kick motion of adolescent soccer players who are experiencing LBP. This study aimed to clarify the kick motion of adolescent soccer players during the presence and absence of LBP.

CLINICAL SIGNIFICANCE

To improve their skills, it is important for soccer players to prevent LBP. We thought that comparing the kick motion of adolescent soccer players with LBP and of those without LBP could clarify the influence of LBP on the kick motion and help identify its causative factors that could help us prevent LBP in adolescent soccer players.

METHODS

We recruited 42 adolescent soccer players and divided them into two groups: presence of LBP group (n = 22) and absence of LBP (NBP; n = 20) group based on the findings of trunk flexion, trunk extension, Kemp's test, and tenderness of the fifth lumbar spinous process.

We measured real-time kick motion using a three-dimensional motion analysis system (Qualisys track manager, Qualisys AB., Sweden). We placed 65 spherical markers on each anatomical landmark and calculated the angle of the lumbar spine, center of mass (COM) of the whole body, and the displacement of the support leg. Based on a previous report [3], we collected data for the following six events: foot contact (FC), toe off (TO), max. hip extension (HE), max. knee flexion (KF), ball impact (BI), and max. hip flexion (HF). We converted the obtained data into 100%. We used unpaired t-test to compare the data between the presence and absence of LBP.

Table 1: Each parameter for both groups

	NBP			LBP			
	mean	±	SD	mean	±	SD	*p*
Age (year)	14.0	±	0.6	13.9	±	0.5	0.801
Height (cm)	164.8	±	8.4	164.1	±	5.5	0.717
Weight (kg)	54.8	±	7.7	52.9	±	6.5	0.368
BMI	20.1	±	1.5	19.6	±	1.6	0.303
Kick time (ms)	513.6	±	128.5	575.2	±	63.7	0.051
Toe off (%)	26.5	±	7.0	27.6	±	7.2	0.594
Hip extension (%)	40.6	±	6.3	42.9	±	6.8	0.263
Knee flexion (%)	57.0	±	5.9	58.0	±	6.0	0.590
Ball impact (%)	69.6	±	6.5	70.5	±	6.3	0.650
Support foot contact (%)	55.1	±	6.9	51.6	±	8.2	0.148

RESULTS

The duration of kick motion for the LBP group was 61.6 ± 30.7 ms greater than that for the NBP group (Table 1). Compared with the NBP group, the LBP group showed a lateral shift in the COM at FC (p = 0.065) and HF (p = 0.076, Fig. 1), and their support foot was positioned posterior from HE (p = 0.078) to BI (p = 0.009). There was no difference in lumbar extension between both the groups. Compared with the NBP group, the LBP group showed 5.9 ± 2.2° greater rotation at FC (p = 0.010) and 3.4 ± 1.8° lesser lateral bending of the lumbar spine at KF (p = 0.067, Fig. 1).

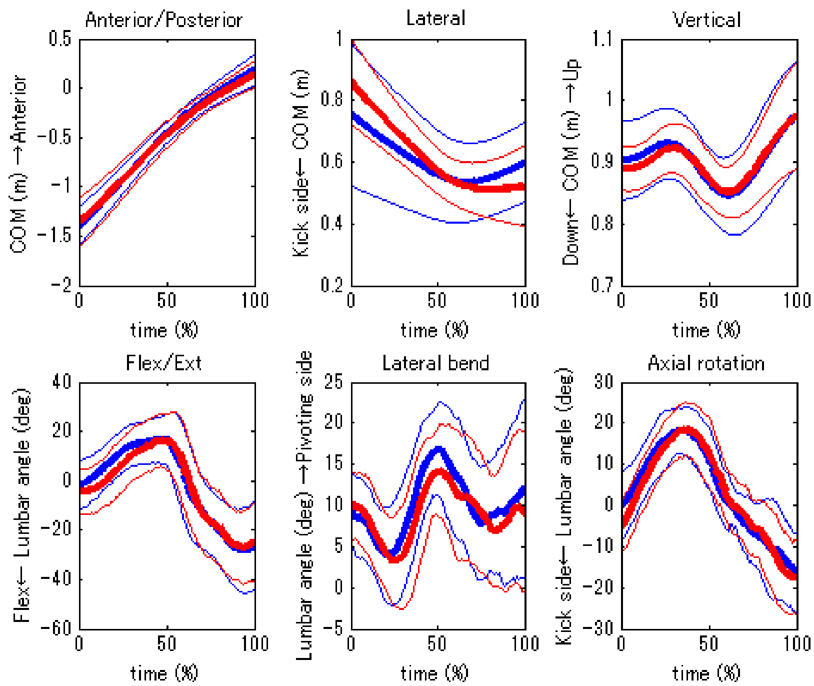

Figure 1: Mean (thick lines) and standard deviation (thin lines) for center of mass (COM) and lumbar spine angle during kick motion. Blue lines, NBP group; red lines, LBP group.

DISCUSSION

Compared with the NBP group, the LBP group showed a lateral shift in the COM. This could increase the duration of kick motion in the LBP group. The presence of LBP could affect the posterior position of the support leg and restrict the player's lumbar spine from bending laterally. Compared with the NBP group, the LBP group showed a lateral shift in the COM, larger rotation of the lumbar spine, and no difference in lumbar extension before BI, which could stress the lumbar spine. Therefore, the coaches and trainers should teach the soccer players to reduce the trunk rotation to the pivoting side at FC and move straight during the kick. This would be important for the adolescent soccer players to prevent LBP.

REFERENCES

[1] Sato T, Ito T, Hirano T, Morita O, Kikuchi R, Endo N, Tanabe N. Low back pain in childhood and adolescence: assessment of sports activities. Eur Spine J 2011; 20(1):94-9.

[2] van der Sluis A, Elferink-Gemser MT, Brink MS, Visscher C. Importance of peak height velocity timing in terms of injuries in talented soccer players. Int J Sports Med 2015; 36(4):327-32.

[3] Brophy RH, Backus SI, Pansy BS, Lyman S, Williams RJ. Lower extremity muscle activation and alignment during the soccer instep and side-foot kicks. J Orthop Sports Phys Ther 2007; 37(5):260-8.

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Effects of body weight support exercise by Spider therapy on the walking of a cerebral palsy patient

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INTRODUCTION

The effect of body-weight-supported exercise has been considered in adult rehabilitation like the body-weight-supported treadmill training. Body-weight-supported exercise is also utilized in cerebral palsy patients, we focused on Spider therapy as a way of improving the mobility of the lower limbs of cerebral palsy patients. The Spider therapy invented by Norman Lozinski in 1993, was administered to a cerebral palsy patient. We suspended the patient's body somewhat upward in four directions with four springs in Spider therapy. It is a part of the Therasuit method in the US. There are few reports about the influence of Spider therapy on gait function. The purpose of this study was to clarify the effect of The Spider therapy on the lower limb joint angle during walking of a cerebral palsy patient.

CLINICAL SIGNIFICANCE

I n this study, we examined the effect of a new therapeutic device that uses stretchable suspension to support the body weight, and contributes to new exercise therapy for cerebral palsy patients.

METHODS

The subject was one cerebral palsy patient(male with spastic diplegia, GMFCSⅡ,21 years old). He was a quadruplet, and was an extremely low-birth-weight infant. He walks with shaking motions, and has a shortened single limb support time. This study was conducted with the approval of the Research Ethics Committee of Suzuka University of Medical Science. The details of the study were explained to the subject, and written consent was obtained.

The subject was suspended with four springs that were stretchable in four directions by the Spider with a weight exemption of 20%. Then, the patient performed the back step and extension-flexion exercise 250 times in total. Gait was measured using a 3D motion analysis system (VICON612) and five force plates (AMTI OR6-6), joint angle and joint moment of the hip, knee and ankle were calculated from spatial coordinates and floor reaction force before and after Spider therapy.

RESULTS

Figure 1 shows the hip joint moment and angle for one walk cycle before and after the

Spider therapy, and Figure 2 shows the knee joint moment and angle.

In this study, after the patient performed the back step in the Spider therapy, the hip extensor moment increased in the early stance phase and the hip maximum extension angle increased in the late stance phase. The maximum extensor moment of the knee decreased at the early stance phase and the maximum extension angle of the knee increased at the stance phase.

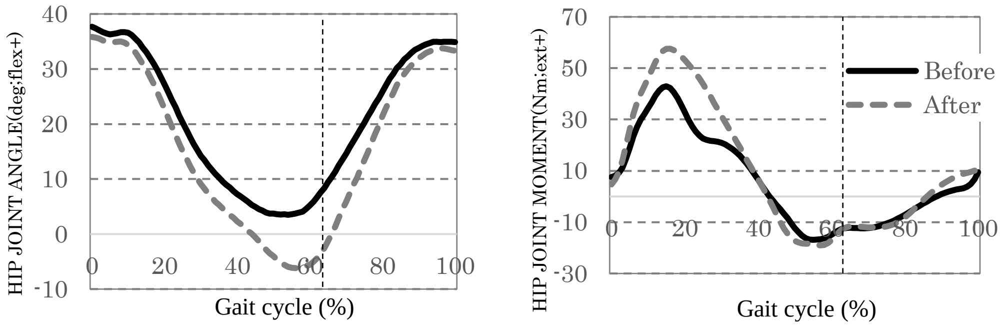

Figure1. Hip joint angle and hip joint moment in one gait cycle before and after therapy.

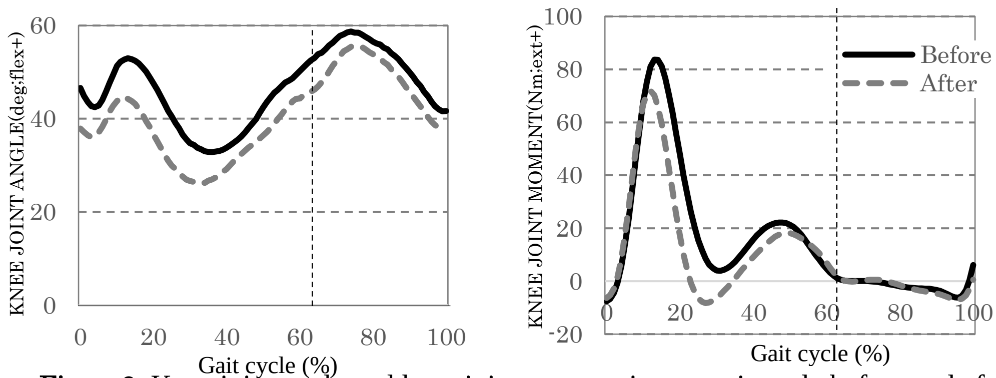

Figure2. Knee joint angle and knee joint moment in one gait cycle before and after

DISCUSSION

The crouch knee gait is one of the four typical abnormal gait patterns seem in cerebral palsy patients. One characteristic of the crouch knee gaits is that the flexion angle of the hip and knee increases during the swing phase. The stretchable suspension spring used in this study provided the movement that supplemented the stability and mobility of the patient, the knee joint was able to expand with a small force, and the joint angle was improved. It is said that the abnormal gait of cerebral palsy patients is difficult to improve, because secondary problems of the musculoskeletal system increase with growth. However, exercise therapy with the body weight supported using stretchable suspension could increase the extension angle of the hip and knee joints in the crouch knee gait of a cerebral palsy patient.

EFFECTS OF VARYING UPPER LIMB USAGE AND RESISTANCE SETTINGS ON MUSCLE ACTIVITY DURING ELLIPTICAL MACHINE E   ERCISE

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INTRODUCTION

Elliptical machine use is an attractive exercise option for many. Users often perceive reduced joint impact loading, relative to overground or treadmill ambulation, as a main advantage of elliptical training. Some elliptical vs walking kinematic similarities, which vary between models, have been described[1]. However, notable kinetic differences have also been reported[2].

Most elliptical machines provide multiple   resistance levels .   hile resistance variations are generally model dependent, users typically perceive increased exercise difficultly as level rises. On many machines, users can grasp either fixed handles or reciprocating bars; or they can choose to grasp neither. Availability, of multiple resistances and three distinct grasping techniques, provides many usage modes. However, potential muscle activity and upper body kinematic differences associated with various elliptical training modes are not fully understood. Consequently, this study sought to understand how varying elliptical machine resistance and grasping technique affects muscle activity and upper body kinematics.

CLINICAL SIGNIFICANCE

Understanding how muscle activity and upper body kinematics are affected by varying elliptical machine resistance level and grasping technique can facilitate formulation of recommendations for how users can adapt elliptical training to address personal needs.

METHODS

Seven healthy subjects (male) participated after giving informed consent. For each subject, 5 of   available pre-amped, surface electrodes were placed over the right anterior tibialis, gastrocnemius, rectus femoris, vastus lateralis & biceps femoris. The other   electrodes were placed over the left gastrocnemius, rectus femoris & biceps femoris. Reflective markers were placed in a Helen Hayes arrangement. For each participant, electromyography (EMG) data were obtained at 1000 Hz with a telemetered system (Delsys), while marker coordinate data were obtained at 100 Hz with a Hawk Motion Tracking System (Motion Analysis Corp.).

All trials were performed on a Precor EF   57 i elliptical machine. For each subject data were obtained for: one 10-second reference (high intensity) trial with resistance at 1 , without grasping, and with maximum cadence; and fifteen 20-second experimental trials with resistance at 1, 4, 7, 10 & 1   for each of the   grasping techniques (fixed handles, reciprocating bars, neither) and with power maintained at 75% of reference trial power.

For each muscle, median RMS windowed EMG cycles (beginning at maximum vertical foot position) were extracted for reference and experimental trials, such that 1^{st} and 2^{nd} half of cycles corresponded to downward and upward elliptical strokes. For each experimental trial, integrations of downward and upward stroke portions of each muscle's median cycle were expressed as percentages of the corresponding muscle's median reference cycle integration. Means of pelvis and trunk angular velocities were computed for each experimental trial.

RESULTS

Left side muscle data predominantly mimicked the right side, and are not further discussed. The gastrocnemius did not exhibit significant grasp or resistance effects. For other muscles, grasp effects were significant for at least one grasp condition during the downward stroke (Fig 1), while resistance effects were significant for at least one resistance level for both downward and upward strokes (Fig 2). Generally, downward stroke values exceeded upward stroke results, except for the anterior tibialis. For pelvis and trunk segment angular velocity means, grasp effects were significant for at least one grasp condition. Only pelvis angular velocity exhibited a significant resistance effect, for at least one resistance level.

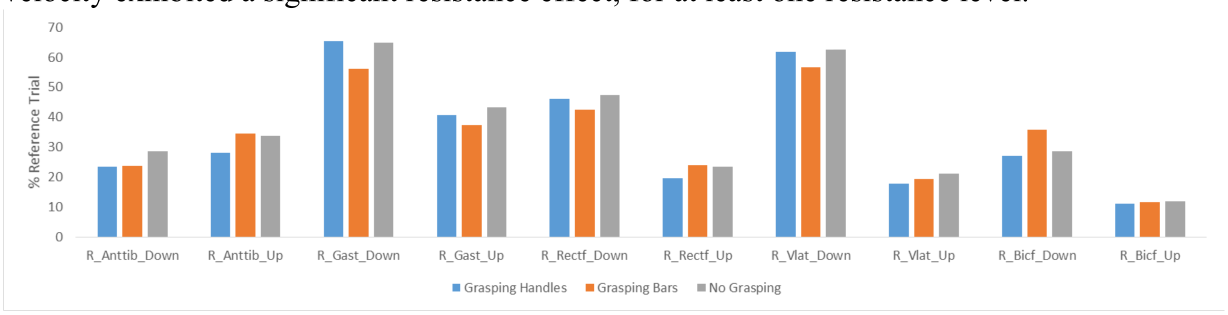

Fig 1. Muscle activities for downward and upward strokes for varied grasping techniques (indicates p 0.05).

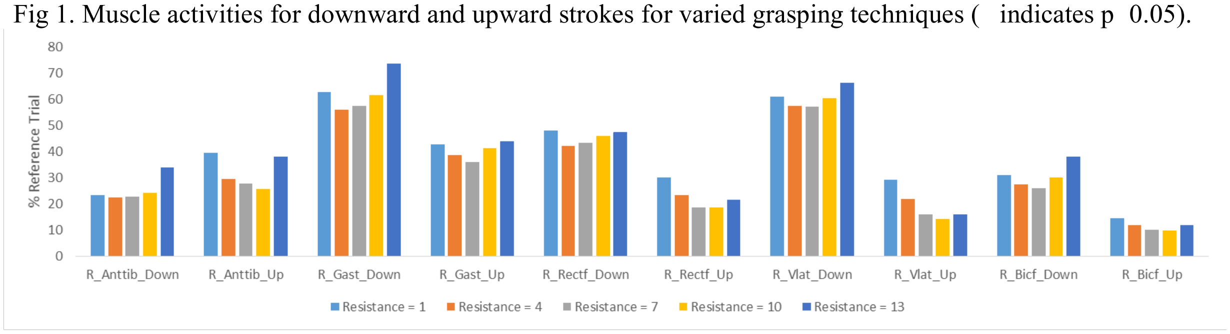

Fig 2. Muscle activities for downward and upward strokes for varied resistance levels (indicates p 0.05).

DISCUSSION

Grasping technique effects on downward stroke muscle activities and pelvis & trunk angular velocities, in conjunction with downward activation levels being generally greater (except anterior tibialis), suggest that motor actuations may ensue to account for posture/stability changes, associated with varying grasping technique. It also appears that both downward and upward resistance effects are associated with presence of local muscle activity minima between very low and high resistance levels. Consequently, achieving similar muscle exertions at mid-range resistances may requires higher cycle frequencies.

REFERENCES

1. Burnfield M et al. (2010) Physical Therapy, 90(2):2 9- 05.
2. Lu T et al. (2007) Medicine & Science in Sports & Exercise, 9(9):1 51- .

DISCLOSURE STATEMENT

The authors have no conflicts of interest to disclose.

FUNCTIONAL OUTCOME COMPARISON BETWEEN 6 AND 12 MONTH FOLLOW-UP FROM ACL RECONSTRUCTION IN PEDIATRICS

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INTRODUCTION: Anterior cruciate ligament (ACL) reconstructions have increased to 18 per 100,000 patients that are under 20 years of age. Currently, the decision of when a patient may return to full activity after surgery is based on subjective measures by the surgeon, physical therapist, and patient, ranging from 6 to 12 months post-surgery. There is no consensus on a functional assessment protocol to assess a patient's recovery or any quantifiable criteria to allow a patient to return to baseline activity. This study aims to: 1) compare the kinematic and kinetic parameters of the reconstructed knee during walking, running, and stair climbing, as well as the isokinetic strength of the lower extremity (LE), and proprioception between 6 and 12-month follow-up time periods to analyze if there is an advantage for patients to delay returning to baseline activities; 2) compare these objective measures between the ACL reconstructed and unaffected LE at 6 and 12 months post-surgery.

CLINICAL SIGNIFICANCE: Understanding the kinematic recovery following a complete reconstruction of the ACL is clinically significant to establish objective criteria to decide when, and if, a patient is safe to return to their baseline activities.

METHODS: Five patients (3 females and 2 males, mean age of 17.0) with a total of 6 ACL reconstructions were analyzed for the 12-month post-op group and four female patients (mean age of 16.8) with a total of 4 ACL reconstructions were analyzed for the 6-month post-op group. Sixteen reflective markers were placed on the LE's. Kinematic and kinetic data was recorded using the T40 Vicon motion system with 12 cameras (Vicon Systems, Oxford, UK) and 4 force plates (Bertec Corp, Columbus, OH and AMTI, Watertown, MA) during walking, running, and stair ascending/descending. Proprioception was compared for each leg with knee flexion to 30 and 60 degrees. Peak flexion and extension moments of the knee were gathered using an isokinetic test on the Biodex system. Vicon's Nexus software constructed a LE model giving 3D kinematics of the hip, knee, and ankle joints for analysis. The mean values between 6 and 12-month follow-up were compared for various parameters using Mann-Whitney U Test. Comparisons of kinetic and kinematic differences between ACL reconstructed and unaffected LE were performed using Wilcoxin Signed Ranked Test.

RESULTS: When comparing the reconstructed to unaffected knee within the 6 and 12-month follow-up groups, there was little significant difference for motion analysis; less flexion/more extension through the hip and knee on the reconstructed side was noted in both follow-up groups. Comparing the 6 and 12-month follow-up groups, significant differences for isokinetic peak torque extension at 180 deg/s and proprioception was seen in the ACL reconstructed knee.

Table 1. Comparison of kinematics and kinetics on the limb with ACL reconstructed knee between 6-months and 12-months post-surgery (Mean ± SD, *P<0.05, % change> ±20%)

Parameter	6 Months Mean ± SD	12 Months Mean ± SD	Percent Change
60-Degree Proprioception	62.78±5.83	52.5±5.1	**-17.6***
Walking Max. Knee Flexor Mz	0.5±0.29	0.36±0.35	-28
Walking Max. Knee Extensor Mz	-0.11±0.11	-0.24±0.2	119.2
Walking Min. Hip Flexion	-16.3±7.9	-10.7±8.4	-34.5
Walking Hip Max. Flexor Mz	0.54±0.37	0.67±0.27	23.7
Walking Hip Max. Extensor Mz	-1.5±0.7	-0.74±0.32	-50.9
Walking Hip Mz at Toe Off	-1±0.88	-0.54±0.32	-46.5
Walking Ankle Min. DF at HC	1.7±6.4	0±1.1	-100.1
Walking Ankle Max.DF at St	18±11	13.1±5	-27.1
Running Max. Knee Flexion at St	31.4±4.3	38.1±7.1	21.3
Running Min. Knee Flexion	9.3±4	16.3±9.8	74.8
Running Max. Hip Flexion at St	25.9±8.4	31.9±8.4	23.3
Running Min. Hip Flexion	-7.1±7.4	0.6±9.1	-108.6
Ascending-Trailing Leg Max Knee Flexion at HC	13.8±8.4	17.7±7.5	28.2
Descending-Lead Leg Max. Knee Flexion at St	13.3±4	18.4±7.3	38.2
Descending-Lead Leg Min. Knee Flexion	3.4±5.4	7.1±6.8	105.7
Peak Torque Extension 300 deg/s	39.9±8.9	52±9	30.3
Peak Torque Extension 180 deg/s	59.8±11.5	81.8±12.1	**36.8***
Peak Torque Flexion 180 deg/s	41.5±12.2	51.3±14	23.9
Peak Torque Extension 60 deg/s	83.7±19.2	116.7±28.9	39.4

Note: Max: Maximum; Min: Minimum; Mz: Moment; “-“: Decreases; “+”: Increases; DF: Dorsiflexion; St: Stance; HC: Heel Contact.

DISCUSSION: Kinematic differences were noted between the 6 and 12-month follow-up groups, notably a significantly higher quadriceps (extensor) peak moment at 12-months post ACL reconstruction ($P<0.05$). Furthermore, there was less flexion and increased flexor moment of the ACL reconstructed knee, as well as increased hip extension and extensor moment during walking at 6 months compared to 12 months ($P>0.05$). Similarly, there was less knee and hip flexion during running, as well as less knee flexion during ascending and descending stairs ($P>0.05$). This data would suggest a compensatory gait exists early after reconstruction and returns closer to baseline as comfort and confidence in the knee progresses. It is also apparent that patients were stronger in the 12-month group versus the 6-month group across all isokinetic measurements for both knee flexion and extension. Given the objective evidence from these biomechanical assessments following the rehabilitative process, motion analysis and Biodex isokinetic measurements may be a useful tool for physicians in determining when a patient may return to full activities. However, it is unclear if these metrics are indeed superior to the current system and further research will need to be done.

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Gait and physical activity of urban-living individuals with Osteogenesis Imperfecta in Southeastern Asia

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INTRODUCTION

Ostogenesis imperfecta (OI) is a rare, highly variable genetic disorder associated with significant impairments to mobility and independence. Activity and weight bearing exercise are thought to be important to bone health, and patients with OI are known to have decreased daily activity compared to healthy controls [1]. Activity can be measured with activity trackers, such as the Fitbit One. Using the Gillette Functional Assessment Questionnaire (FAQ), self-reported barriers to mobility include pain, weakness and walking ability and we examined how these correlated with gait and daily activity as well as the feasibility of using the Fitbit and gait analysis to measure function.

CLINICAL SIGNIFICANCE

The Gillette FAQ Walking Scale is highly indicative of daily activity of participants with OI, and self-reported weakness on the Gillette correlates with gait function in those with OI. The Fitbit is a novel way to track activity levels and mobility in OI.

METHODS

We evaluated 33 participants (age 13.7±15.6, 11 males and 22 females) with OI in a cross-sectional study. Of the initial 33 patients (19 Type I/IV/V, 6 Type III, 8 unknown), 16 (12 Type I/IV/V, 6 Type III) were independently ambulatory and were enrolled in the study. Participants answered the Gillette FAQ and then underwent gait analysis for assessment of kinematics, walking speed, stride length and cadence. The first ten participants (8 Type I/IV/V, 2 Type III) went home with Fitbit One activity trackers for one week to acquire daily step counts.

RESULTS:

We found that the daily step counts were typically below previously reported values for typically developing individuals, and step counts varied widely within each group (Table I). Participants with severe OI (type III) had decreased daily step counts compared to mild-to-moderate OI (types I, IV, V), but our sample size was too low to determine statistical significance. On motion analysis, the OI participants had decreased walking speed and cadence than historical controls. The Gillette FAQ had multiple measures that were predictive of participant Fitbit activity level and gait characteristics. The Gillette Walking Scale score was highly predictive of Fitbit daily step counts ($p=0.02$). Patient-reported weakness on the Gillette FAQ was predictive of decreased walking speed ($p=0.022$), and decreased cadence ($p=0.016$).

Table 1: Daily Activity Levels (recorded using Fitbit One activity tracker)

OI Type	N	Age (yrs)	Gillette Walking Score	Daily Steps (weekday)	Daily Steps (weekend)	% Sedentary (weekday)	% moderate to vigorous (weekday)
Mild-to moderate (types I, IV, V)	8	22.9±23.3	45.0±18.7	6,070±3666	6596±4706	83.04±7.64%	0.64±0.62% (27 minutes)
Severe (type III) #1	1	18	71	392	78	95.78%	0.00%
Severe (type III) #2	1	19	53	4064	3521	88.76%	0.54%
All Participants	10	22.0±20.7	45.0±18.7	5217	5530	85.05%	0.56%
Typically Developing [2]	*14*	*12.75±4.62*	-	*9463*	*7058*	*56.11%*	*5.31%*

Table 2: Gait Parameters (recorded using laboratory gait analysis)

OI Type	N	Age (yrs)	Gillette Weakness Score	Gillette Walking Score	Walking Speed (m/s)	L Steps/min	R steps/min
Mild-to moderate (I, IV, V)	14	22.7±21.1	1.5±0.7	51.2±22.0	0.71±0.23	111.14±20.8	108.92±19.6
Type III #1	1	18	1	71	0.11	19.12	19.29
Type III #2	1	19	2	53	0.81	55.58	54.89
All Participants	16	21.0±19.4	1.5±0.7	51.7±20.5	0.67±0.26	108.46±27.9	106.11±26.7
Historical Typically Developing [5]	22	10.3±3.1	-	-	1.2±0.0	122.9±2.8 (combined)	122.9±2.8 (combined)

DISCUSSION

There is high variability in activity levels among participants with OI, though they are less active than their healthy counterparts. Self-reported weakness appears to be a predictor of slower gait in a laboratory setting. Our results suggest that rehabilitation must focus on weakness, and that self-reported measures are reliable indicators of activity and daily function.

DISCLOSURE STATEMENT

The authors have no conflicts of interest to disclose.

REFERENCES

1. Pouliot-Laforte et al. (2015) Physical activity in youth with osteogenesis imperfecta type I

2. Brizola et al. (2014) Muscle Strength, Joint Range of Motion, and Gait in Children and Adolescents With Osteogenesis Imperfecta

3. Graf et al. (2009) Gait characteristics and functional assessment of children with type I osteogenesis imperfecta

Immediate effects and aftereffects of using a gait assist suit device when walking without the device

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INTRODUCTION

In gait exercise, it is important to consider repetition and task-specific effects based on motor learning theory. A gait exercise support robot can be used to maximize repetition and task-specific effects[1].

One walking support machine with added assist function using robot engineering technology is the non-powered gait assist suit device (NP-GAD). As with other gait exercise support robots, even in gait exercises using NP-GAD, effect decision and the best usage methods are being developed.

Therefore, in order to maximize the effect of gait exercise using NP-GAD, we investigated the influence of assist by NP-GAD on walking in healthy subjects. In addition, we examined the immediate effect of gait exercise using NP-GAD in regards to repetition and task-specificity. However, it remains unclear whether the effect is immediate or an aftereffect.

The aim of the present study was to clarify the effect of gait exercise using NP-GAD on the gait pattern of healthy subjects after 1 hour.

CLINICAL SIGNIFICANCE

The results of this study can help maximize gait exercise effect with power assist function using robot engineering technology.

METHODS

Twenty-three healthy adult volunteers participated in this study. All subjects were fitted with NP-GAD and underwent gait exercise of 1000 steps (500 steps for each leg), as described previously[2]. Data were collected while barefoot walking before gait exercise (before ex), walking immediately after gait exercise (after ex), and walking at 1 hour after gait exercise (1hr). In barefoot walking and gait exercise, in order to avoid the influence of gait velocity, the step length was made constant at 45% of the participant height and with the walking rate set to 120steps/minute using a metronome. Data were collected after sufficient practice. Subjects walked on the walking path with a tape affixed in 45% increments of participant height.

Gait was measured using a 3D motion analysis system (VICON612) and 5force plates (AMTI: OR6-6). The joint moments of the hip, knee, and ankle were calculated using the inverse dynamic model. From these data, the peak values of joint angles and joint moments were extracted and compared between before ex, after ex, and 1 hr. The kinematics and kinetics data in the 3 conditions were assessed using one-way ANOVA, with post-hoc comparison using the Bonferroni adjustment. Statistical significance level was less than 5%.

This study was approved by the Ethics Committee of Suzuka University of Medical Science and was performed in accordance with the Declaration of Helsinki. Informed consent was obtained from all participants prior to their participation in this study.

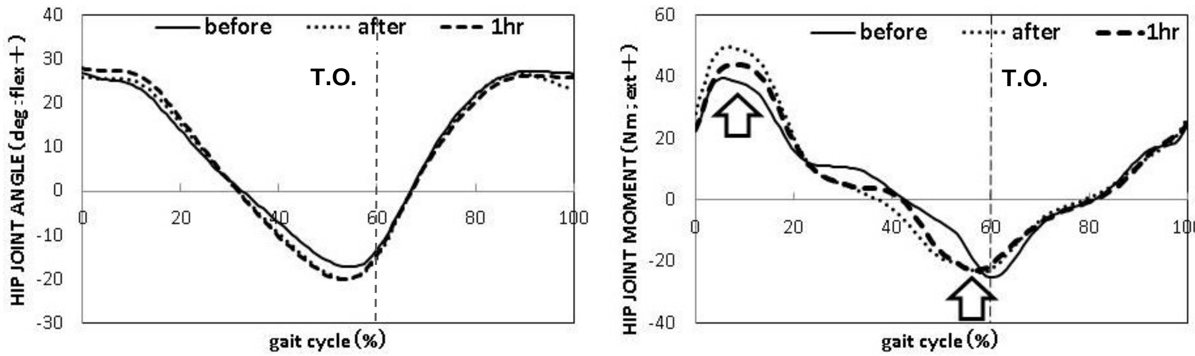

Figure1. Typical examples of hip joint angle and hip joint moment of the three conditions over one gait cycle.

RESULTS

Figure 1 shows a typical example of the hip joint angle change and joint moment change during one gait cycle.

Characteristic points of extensor moment in the loading response, hip joint extensor moment in the mid-stance, and maximum flexor moment in the pre-swing phase of the hip showed significant differences. Specifically, hip joint extensor moment in the loading response (p <0.01) and maximum flexor moment in the pre-swing phase (p <0.01) of the hip showed significant differences when comparing after ex and 1hr to that of before ex. In particular after 1hr, the maximum extension moment of the loading response phase increased by 5.4 Nm and the maximum hip flexion moment decreased by 3.5 Nm compared to that observed before ex.

Table 1: The difference in hip joint moment parameters between the three conditions [mean±SD].

HIP JOINT MOMENT [Nm]					
phase	**before ex**	**after ex**	**1hr**	**before vs after**	**before vs1hr**
LR(ext moment)	**31.9±10.2**	**38.7±11.8**	**37.2±10.7**	**	**
Mst(ext moment)	**21.9±10.9**	**27.0±12.7**	**22.9±12.0**	**	n.s.
Psw(flex moment)	**31.9±8.5**	**27.7±8.6**	**28.5±8.0**	**	**

［**: p ＜0.01］

DISCUSSION

We considered anticipatory postural adjustments (APAs) for assist by NP-GAD as factors for the effect lasting until 1hr after gait exercise. APAs are motor responses generated in a feedforward manner by the central nervous system to stabilize balance prior to expected internal perturbations caused by fast voluntary movements[3)]. It was considered that aftereffects were acquired because gait exercises using NP-GAD were repeatedly practiced feedforward postural responses. Our results demonstrate that gait exercise using NP-GAD influence the gait pattern even after 1hr due to repetition and task-specific effects.

REFERENCES

1. Sale, P. (2012) Eur J Phys Rehabil Med, pp.111-121.
2. YIN, Ping-Bo. (2001) Experimental Brain Research, pp.250-253.
3. S. Shiozawa（2015）European Journal of Pain, pp.1362-1371.

KINEMATIC DIFFERENCES BETWEEN PLUG-IN-GAIT AND A 6-DEGREE-OF-FREEDOM MODEL DURING WALKING AND CUTTING

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INTRODUCTION

A variety of different models can be used to estimate joint kinematics from marker-based motion capture data. The Plug-in-Gait (PiG) model is most commonly used for traditional clinical gait analysis. The defining characteristic of this model is that each segment is linked to its parent, stemming from the global pelvis. The links between segments impose constraints on segment motion in the model, therefore having the potential to overlook actual motion. Some newer research in gait and sports studies has adopted models where each segment is independent with no imposed links between segments. This type of model is generally called a 6-degree-of-freedom (6DoF) model, though there are many variations in methodology. The purpose of this study was to quantify the difference between joint kinematics produced by the PiG model and a 6DoF-type model.

CLINICAL SIGNIFICANCE

For reasons of universal comparability it is important to understand the differences between models, especially any systematic differences in kinematic results. For assessment of patients, the differences in kinematic outputs must be considered when comparing to reference data and interpreting results.

METHODS

Motion of 27 subjects (average age 13.5 ± 1.8yrs; 52% female) was recorded during walking, cutting, and other sports tasks using a 9-camera Vicon motion capture system. Subjects simultaneously wore enough reflective markers to process both models. The standard PiG marker set was used with an anterior patella marker replacing the standard thigh marker (Figure 1a). The 6DoF marker set utilized 6 markers for calibration only and three or four markers unique to each segment for tracking (Figure 1b). The PiG data was labelled, gap filled, filtered, and processed in Vicon Nexus. The 6DoF model was labelled, gap filled, and filtered in Vicon Nexus and processed in Visual3D.

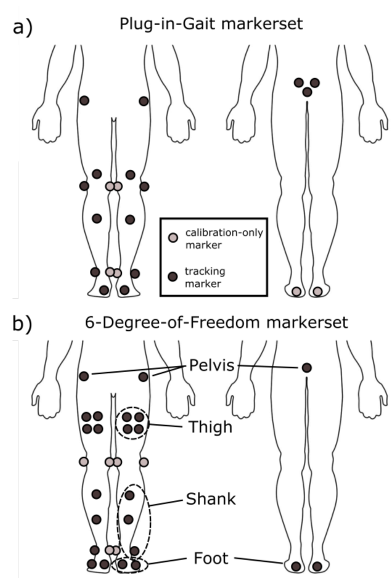

Figure 1. PiG (a) and 6DoF (b) markersets.

The PiG and 6DoF kinematic curves for all subjects were averaged together and normalized by gait cycle for walking and foot contact phase for cutting. Normalized difference curves were created by taking the differences between the average PiG and 6DoF model curves of the hip, knee, and ankle during walking and cutting. As a metric of overall difference, the area under each kinematic difference curve

(AUC) was calculated by integrating the absolute value of each difference curve.

RESULTS

For both walking (Figures a, c) and cutting (Figures b, d), the greatest overall (AUC) difference was in hip abduction/adduction, with the 6DoF model indicating greater adduction. The three knee angles also showed large differences, including fluctuations between positive and negative differences. The overall AUC difference was greater in the cut compared to walk in all four of these angles.

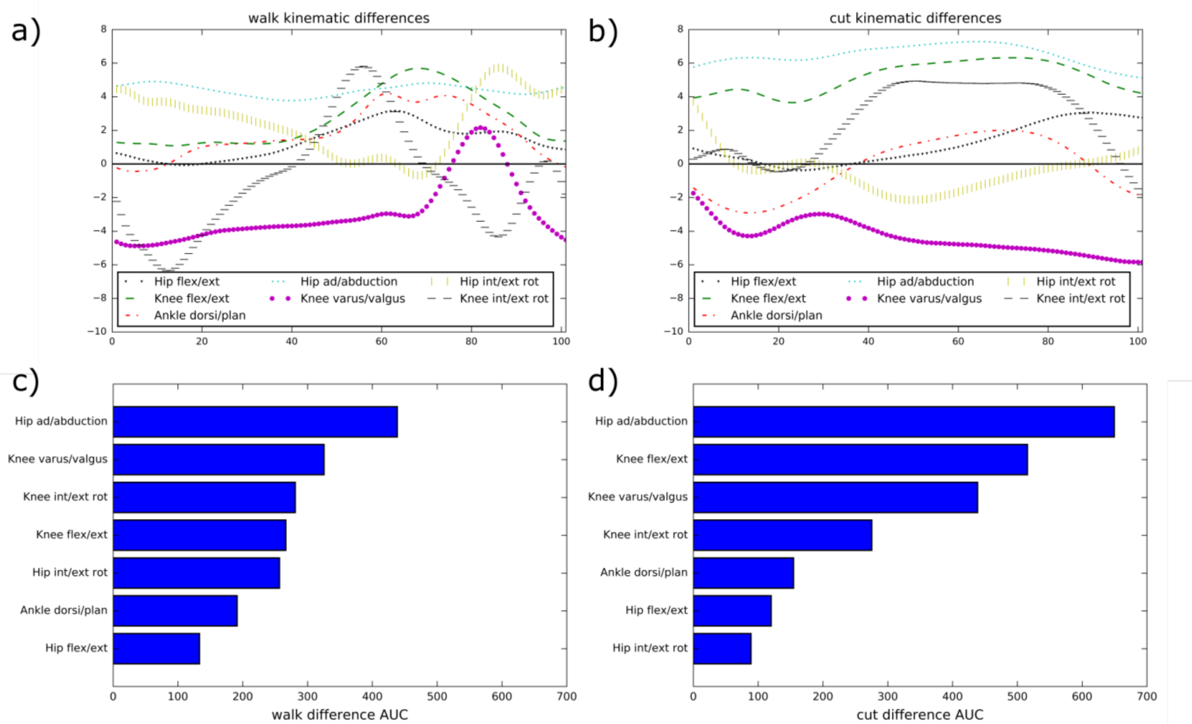

Figure 2. The differences between the PiG and 6DoF model kinematic curves of the hip, knee, and ankle during walking (a) and cutting (b). The area under the curve (AUC) of each difference line, in descending order for walking (c) and cutting (d).

DISCUSSION

These findings suggest systematic differences in kinematics resulting from the PiG and a 6DoF model. The magnitude of difference is dependent on the activity being measured. These findings illustrate the importance of considering the activities being evaluated when choosing a model for clinical or research purposes. It is also important to understand model differences when interpreting and comparing with data from the literature. Finally, it is essential to fully describe the model used when disseminating research to others.

ACKNOWLEDGMENTS

We would like to thank our team in the Motion Lab – Henry Lopez, Mia Katzel, Bitte Healy.

DISCLOSURE STATEMENT

None of the authors have conflicts of interest to disclose.

LONGITUDINAL INSTRUMENTED GAIT ANALYSIS ON THE USE OF A CARBON FIBER AFO FOR A CHILD WITH CEREBRAL PALSY: CLINICAL CASE STUDY

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PATIENT HISTORY

A 13-year-old male with spastic diplegic cerebral palsy (CP), GMFCS level I, presented to the Motion Analysis Laboratory for a two-year post-operative instrumented gait analysis (IGA) to assess effects of open bilateral adductor tenotomies and evaluate patient for further treatment intervention. Gait study results indicated presence of bilateral forefoot adduction, midfoot supination, hindfoot varus, decreased knee motion, and weakened gastroc-soleus muscles. Bilateral carbon fiber AFOs with SMO inner boots (CF-AFO/SMOs) were recommended as a treatment intervention to address gait deficiencies. The goals of the carbon fiber AFO component of this orthotic design were to improve knee motion and increase plantar flexor strength. The goals of the SMO inner boot component of this orthotic design were to maintain and improve bilateral ankle/foot alignment. After receiving the recommended bilateral CF-AFO/SMOs, a longitudinal series of three IGA studies were performed at the following intervals of orthotic use: Day One, Six months, and Twelve months.

CLINICAL AND MOTION DATA

The Day One gait study was performed immediately after patient was fit with bilateral CF-AFO/SMOs. Four walking conditions were collected and compared: Barefoot, Sneakers only, SMOs only, and CF-AFO/SMOs. Comparison of the IGA data for each walking condition yielded significant improvement in knee motion with increasing peak knee flexion and increased total knee motion during gait with use of bilateral CF-AFO/SMOs as compared to other walking conditions. Subject also presented with improved pelvic alignment in the transverse plane and improved temporal-spatial parameters with CF-AFO/SMOs.

The Six-Month gait study was performed to assess short term effects of CF-AFO/SMOs on gait and function. Two walking conditions were examined, Barefoot and CF-AFO/SMOs. Comparison of the IGA data demonstrated improved knee motion with CF-AFO/SMOs as compared to barefoot. Comparison of Day One vs. Six Month Barefoot conditions also demonstrated improvement of knee motion supporting possible carry over effects when AFOs are not donned. Subject's gait (measured by Gait Deviation Index) improved in left/right symmetry with and without orthoses. Parents reported a noticeable reduction in frequency of falling with current orthotic intervention. Parents and patient reported happiness with CF-AFO/SMOs and believed patient's walking and running had improved over the last six months.

The Twelve-Month gait study was performed to assess longer term effects of CF-AFO/SMOs on gait, function, and clinical exam of the patient. The following three walking conditions were examined: Barefoot, CF-AFO/SMOs, and CF-AFO/SMOs with a ¼ inch heel wedge. Continued knee motion improvements were noted. A longitudinal comparison of the subject's clinical exam yielded improvement in plantar flexion strength, approximately two muscle grades. During periods of rapid growth, it is typical for a child with CP to have a decline in strength and function. The noted improvements in plantar flexion strength, gait, and function for this 13-year-old male with CP are significant given the patient grew 4.3 inches over study year.

TREATMENT DECISIONS AND INDICATIONS

The bilateral CF-AFO/SMOs recommended to this 13-year-old male with spastic diplegic CP functioning at GMFCS level I appear to have meet the orthotic goals of improved knee motion and increased plantar flexor strength both in short-term as well as long-term assessment.

SUMMARY

This longitudinal, single subject IGA case series offers objective motion and clinical exam data that supports the use of bilateral CF-AFO/SMOs in a subject with spastic CP functioning at a GMFCS level I. Further IGA research is needed to advance the understanding of effects of carbon-fiber AFOs with varying orthotic set-ups (ex. varying heel wedge heights, presence of SMO inner boots) on gait, function, and clinical exam to advance orthotic prescription and, therefore, patient care.

DISCLOSURE STATEMENT

C. Bickley, J. Linton, and D. Barnes have no conflicts of interest to disclose.

Muscle Activation Pattern during Gait and Stair Activities following Total Hip Arthroplasty with a Direct Anterior Approach: a Comprehensive Case Study

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PATIENT HISTORY

A 60 year old male, BMI 30.4, with symptoms of right hip pain and stiffness is presented.

CLINICAL DATA

Radiographic reports showed bilateral advanced hip arthritis with bone-on-bone changes. The left hip displayed no symptoms. The patient reported no other joint or chronic health problems. The patient's preoperative Harris hip score was 64/100.

MOTION DATA

Comprehensive bilateral gait and stair assessments were carried out preoperatively as well as 3 and 12 months postoperatively. Gait assessment methods include electromyography (EMG) and lab based motion analysis using infrared cameras and force plates. EMG data was collected from 7 major muscles including the gluteus maximus (GMax), gluteus medius (GMed), iliopsoas (Iliop), tensor fascia lata (TFL), rectus femoris (RF), hamstrings (HM) and the lumbar erector spinae (Lumbar ES). Both time and frequency domain EMG were assessed during level walking and stair climbing.

TREATMENT DECISIONS AND INDICATIONS

Total hip arthroplasty (THA) with a direct anterior approach (DAA) was performed on the right hip. Postoperatively, the patient was trained to use crutches and a cane and was asked to bear as much weight as tolerable on the operated leg. The patient was discharged two days after the surgery and no further physical therapy was prescribed.

OUTCOME

The patient's walking speed returned to normal, increasing from 1.17 m/s preoperatively, to 1.3 m/s at 12 months postoperatively. The Harris hip score also improved, recording 100/100, representing an overall improvement. Joint power and recorded moments started to show improvement from 3 months onwards (Fig. 1). Looking at the time domain, EMG revealed an abnormal activation pattern, including aphasic and/or co-contraction on both sides or on the left asymptomatic side (adapted gait pattern) in all assessed muscles. Frequency domain EMG displayed a left spectral shift of median power frequency (MdPF) at 12 months postoperatively, for GMax, Iliop, and lumbar ES with MdPF <70 Hz (Fig. 2). In lower limb muscles, an MdPF of ≤ 70 Hz is considered as a sign of fatigue, representing a dystonic behavior [1].

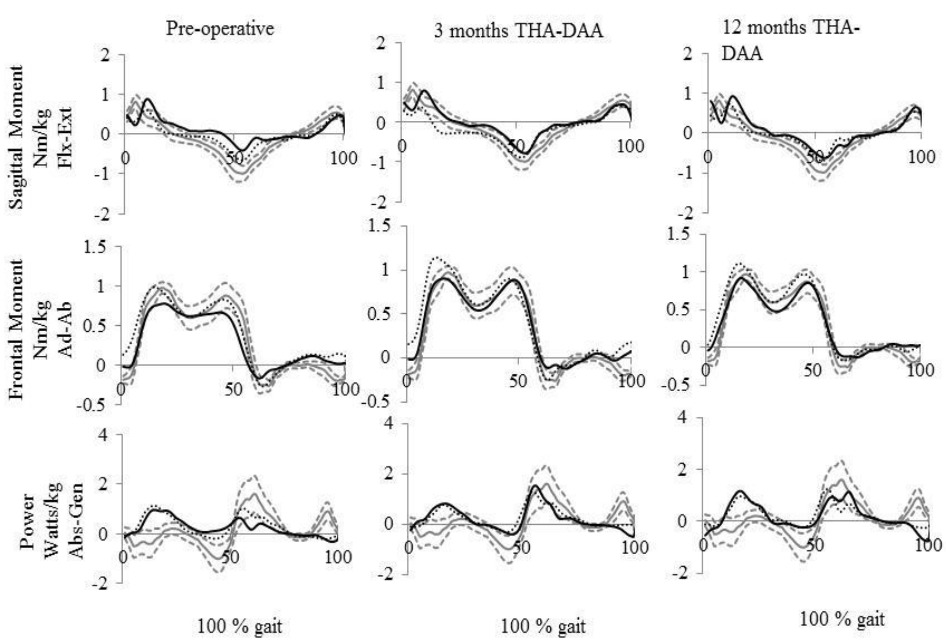

Figure 1: Moment and Power outcome over 100% of the gait cycle. Grey solid and dashed line represents means and SD in controls, black solid line represents the operated side, and black dotted line represents the contralateral side.

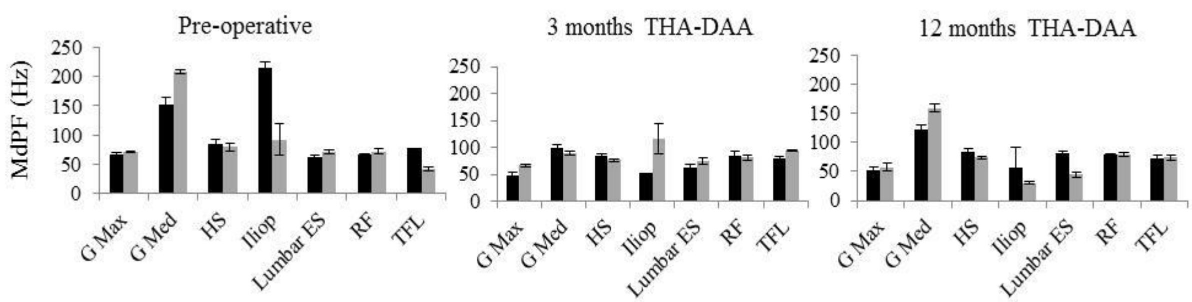

Figure 2: Frequency domain EMG outcome for operated (black) and contralateral asymptomatic (grey) side, MdPF (median power frequency).

SUMMARY

Different surgical approaches are used to perform THA, with DAA being considered the most economical as it does not require muscles to be transected, leading to a faster recovery. However, up to 12 months following DAA, most muscle function were not seen to return to the normative range, even with a good clinical outcome. As a result, although DAA has shown a faster short-term recovery, over time there is no difference in the outcome between various approaches. This case study therefore concludes that DAA-THA may well have the potential to be the superior approach in terms of muscle recovery, but the patient may benefit from a postoperative rehabilitation.

REFERENCES

1. Go, S.A. et al, (2014), Journal of Electromyography Kinesiology.

DISCLOSURE STATEMENT

No conflicts of interest to disclose.

Recovery after Single Event Multi-level Lower Extremity Surgery: Pilot Study

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INTRODUCTION
Single event multi-level surgery (SEMLS) to improve gait quality is often warranted for children with cerebral palsy (CP) to optimize functional outcomes. Post-operative rehabilitation is vital to recovery. This pilot study evaluated whether functional recovery time at 12 months post-surgery was reasonable for youth with CP and whether functional gait improvements can be seen in that time frame.

CLNICAL SIGNIFICANCE
It is important to be able to provide patients and families with realistic expectations for post-operative recovery and gait outcomes for SEMLS.

METHODS
Nine participants (average age 11y 3m; 6M, 3F; 7 diplegia, 2 triplegia; 5 Gross motor Functional Classification System (GMFCS) II, 4 GMFCS III pre-operatively) who had SEMLS surgery between 2012 and 2015 were seen for motion analysis pre-operatively and at 12 months. Each of the children participated in a 2-4 week intensive rehabilitation admission 4-8 weeks after surgery with twice daily physical therapy treatments including active/resistive lower extremity exercises, partial body-weight support treadmill training, over ground gait training, aquatic therapy, and performance of functional activities (bed mobility, transfers, transitions and stair negotiation). Outcomes included lower extremity gait kinematics, Gait Deviation Index (GDI), Functional Mobility Scale (FMS), and parent reported Pediatric Outcomes Data Collection Instrument (PODCI).

RESULTS
Outcomes at 12 months post-operative are detailed in Table 1. At 12 months post-operative, GMFCS levels were unchanged except for one subject who went from GMFCS II to III. Significant improvements from pre-operative to 12 months post-operative were seen for: single support time and GDI for the left side. Significant decreases from pre-operative to 12 months post-operative were seen for: cadence, double support time, stride length, walking speed, and 500m FMS score. There were no significant changes in parent reported PODCI scores.

DISCUSSION
A standardized intensive post-operative therapy program post SEMLS led to improvements in GDI scores and single support time at 12 months post-operative for our patients at various GMFCS levels. Based on limited previous studies, we anticipate that our patients would improve even further at 2 years post-op follow up. Results from this case series can help provide patients and families with realistic expectations for post-operative recovery, realizing that the patients often don't return to their baseline or higher function even at one year post-operative. Future studies should look at outcomes at 2 years and beyond to determine the true benefit of surgery.

ACKNOWLEDGEMENTS

We would like to acknowledge the Helen Kay Charitable Trust grant for funding this research.

DISCLOSURE STATEMENT
None of the authors have conflicts of interest to disclose.

Table 1. Pre and post-operative outcome comparisons.

Parameters	Mean Pre-op	Mean at 1 Year	P-value	Typically Developing Scores
Cadence average R/L (steps/min)	67.81	55.62ǂ	0.034	125±15.1 steps/min
Double support average R/L (% cycle)	28.84	33.49ǂ	0.002	22.0± 2.99%
Single support average R/L (% cycle)	35.14	32.93ǂ	0.010	39.1±2.35%
Stride length average R/L (m)	0.85	0.78 ǂ	0.009	1.14±0.16m
Walking speed average R/L (m/s)	0.84	0.65ǂ	0.005	1.17±0.100m/s
PODCI transfers and basic mobility (100 max)	67.71	66.71	0.839	99.7 (0.9)
PODCI sports and physical function (100 max)	42.29	37.14	0.529	95.4 (8.2)
PODCI comfort and pain (100 max)	57.29	65.43	0.557	91.1 (13.9)
PODCI global function (100 max)	59.71	61.14	0.815	96.4 (5.5)
FMS score for 5 meters (Scale 1-6 max)	4.78	4.0	0.065	6
FMS score for 50 meters (Scale 1-6 max)	3.56	3.44	0.594	6
FMS score for 500 meters (Scale 1-6 max)	3.33	2.33ǂ	0.053	6
GDI Indices Left	52.38	65.75ǂ	0.034	100
GDI Indices Right	56.75	63.00	0.108	100

ǂ Significant at 0.05 level using 2-tailed paired samples test pre vs post-operatively

Relationship of Anthropometrics and Throwing Biomechanics in Youth Baseball Pitchers

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INTRODUCTION

Youth athletes present with a particular subset of conditions and injuries such as Little League elbow, Little League shoulder, osteochondritis dissecans, Panner's disease and multidirectional instability of the shoulder [1] that are related to anatomical and developmental variances including open physes, joint laxity, and strength deficits. Many of these throwing related injuries are due to the accumulation of micro-trauma experienced from repetitive throwing motions [2].

Correlations between anthropometrics and kinematic/kinetic evaluations of the shoulder and elbow of youth baseball player pitching motions have not been well defined. The purpose of this study was: 1) to determine the correlation between anthropometrics including age, body mass index (BMI), hand size, and grip strength and throwing biomechanics in youth baseball pitchers; 2) to determine which characteristics may be predictive of throwing biomechanics.

CLINICAL SIGNIFICANCE

This study can provide useful data for understanding the biomechanical differences in youth pitchers, which may predict who will be prone to alteration of motion of the shoulder and elbow leading to injuries of the upper extremity.

METHODS

Twenty-six healthy male subjects between ages 9-14 years old were recruited from the Milwaukee area youth baseball clubs (mean age: 11.8 years; height: 156.1 cm; weight: 45.6 kg). Anthropometric measurements were taken manually. Grip strength was measured using Jamar Dynamometer. Three max effort reps at anatomical position were averaged and reported.

For kinematic analysis, 44 reflective markers were placed on the body according to the full-body plug-in-gait model. After warming up, subjects were asked to throw ten of their most accurate fast balls with maximal effort into a net twelve feet from the constructed pitching mound and pitching motion was recorded with twelve T40-S Vicon motion capture cameras. Three pitches were used for kinematic data processing. A segmented biomechanical model was created allowing kinematic measurements of the shoulder in three-dimensional planes and movements of the elbow in the sagittal plane. Kinematic data was analyzed through three different phases involving the late-cocking, acceleration, and deceleration phase. Isokinetic and isometric external rotation (ER) and internal rotation (IR) shoulder strength was assessed on the Biodex. Isokinetic shoulder strength was tested at 60°/sec and at 180°/sec. Isometric shoulder strength was tested at 0°, 45° and 105°. Max torque generated from each test was recorded.

Spearman correlation between a single anthropometric measurement and kinematics and kinetics were calculated using a linear regression model. Combined variables from anthropometrics were selected to predict kinematics and kinetics using a parsimonious model. $P<0.01$ was considered significant.

RESULTS

There were no significant correlations between age and kinematics of UE, while the rest of the anthropometrics showed significant relationships with shoulder adduction or abduction at late cocking and deceleration phase as well as with elbow ROM at deceleration phase (see table 1). Kinetic analysis showed there was a significant correlation between age, BMI, hand size and grip strength with peak torque at shoulder ER, especially grip strength having significant relationships with both isometric and isokinetic external rotation torque (r ranging from 0.49 to 0.72). BMI or hand size coupled with grip strength significantly correlated with shoulder rotations in the coronal or transversal plane at deceleration or late cocking phase ($P<0.01$). Hand size combined with grip strength or age significantly predicted isokinetic and isometric peak torque at external rotation.

Table 1. Significant correlations of shoulder or elbow kinematics and kinetics with anthropometrics (Spearman correlation coefficient (r), $P<0.01$)

Parameters	Age	BMI	Hand Size	Grip Strength
Kinematics:				
Min. Deceleration. Coronal. Shoulder			-0.425	
Late Cocking. ROM. Coronal. Shoulder		0.551		0.468
Deceleration. ROM. Coronal. Shoulder			0.575	
Deceleration. ROM. Sagittal. Elbow		-0.622		
Kinetics:				
Isokinetic ER 60°/sec	0.610	0.604		0.718
Isokinetic ER 180°/sec	0.526	0.604	0.473	0.568
Isometric ER 0°			0.551	0.491
Isometric ER 45°	0.661			0.717
Isometric ER 105°	0.567	0.541	0.486	0.514

DISCUSSION

A previous study showed that age, BMI and grip strength are significant predictors of the ball kinetics thrown by youth baseball players [3]. The torque produced by internal rotator muscles at the shoulder during the pitching motion is opposed by the external rotators of the shoulder. Therefore highlighting the external rotators protective role in deceleration and preventing anterior joint impingement and labral tears at the shoulder [4]. Our findings support that these anthropometric variables may be used as strength and motion predictors to prevent injury.

REFERENCES

1. Tisano, B. K. et al. (2016) American College of Sports Medicine 48(10):1898-1905.
2. Lyman, S. et al. (2002) The American Journal of Sports Medicine, Vol. 30, No. 4.
3. Nakata, H. et al. (2013) Journal of Strength and Condition Research 27(10):2887–97.
4. Greiwe, M. R. et al. (2010) Orthopedic Clinics of North America; 41: 309-323.

DISCLOSURE STATEMENT

No conflicts of interest to disclose.

RELIABILITY ASSESSMENT OF GAIT KINEMATICS IN PEDIATRICS: A SINGLE CASE STUDY

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INTRODUCTION

The variability of adult gait has shown to be relatively high within and between sessions of measurements [1, 2]. Gait is more consistent in adults than in children [3] and given the fact that children have smaller segments and bones, an inconsistency in an examiner's measurement may influence kinematic outcomes; hence the variability reported from adults might not be generalized. This study examined the variability of gait parameters in two similarly-equipped gait analysis laboratories to determine the effects of subject and measurer-related inconsistencies in pediatrics.

CLINICAL SIGNIFICANCE

The general variability was acceptable but it is noteworthy to mention that small differences in marker placement caused an effect on some kinematic measures of the participant while difference in anthropometric data did not change the results.

METHODS

Gait analysis was performed on a typically developing 4-year-old female child in a Shriners hospital for children at Shreveport (Lab 1) and the Center for Pediatric Locomotion Sciences, Georgia State University (Lab 2) within a period of two weeks. The protocol was approved by the ethical committees at each site, and the subject's parent provided consent. Laboratories utilized identical Vicon Motion Analysis system (Vicon Motion Systems Ltd. UK). Examiners were provided with a standardized gait analysis protocol training video (Shriners Hospitals for Children). Each examiner collected anthropometric data and attached sixteen reflective markers on the anatomical landmarks according to the Vicon Plug-in-Gait marker set manual. The subject walked barefoot at a comfortable, self-selected walking speed and kinematic data were collected at the frequency of 100 Hz. The Plug-In-Gait Lower Limb model was used for lower limb joint angle calculation and plots were generated using Polygon software. The average and Standard Deviation (S.D.) values for each joint angle were calculated. Joint kinematics of lab 1 were also recalculated using the anthropometric data of lab 2 to assess the effect of inconsistent anthropometric measurements.

RESULTS

Anthropometric measurements differed by no more than 8% at either site. Average walking speeds and stride lengths were 0.90m/s, 1.08 m/s and 0.79m and 0.84 m for Lab 1 and Lab 2, respectively. At both labs, pelvic obliquity was the most reliable for lab 1 and knee flex/ext angles indicated the largest error (Table 1). Except for the pelvic joint, Lab 1 showed relatively larger variability (mean S.D. = 2.52°) than lab 2 (mean S.D. = 2.0°). With the adjusted anthropometric data, average variability for lab one reduced to 2.49°.

Figure 1 represents the average angle graphs for lab 1, 2 and lab 1 with adjusted anthropometric measures. A consistent shift in the curves was seen even with the adjusted anthropometric measures.

DISCUSSION

Walking speed was slower and strides were shorter in lab 1. Peak ankle plantarflexion was 7° greater in Lab 2, where the subject walked faster. Gait patterns are more variable in children, and may stabilize at or after skeletal maturity [3]. Therefore, differences in pediatric walking speeds should be considered a source of intrinsic variability when studying joint kinematics in this population.

In the Plug-in-Gait model, the position of pelvis segment is defined by the ASIS and PSIS markers. Knee flex/ext axis is determined by the hip joint center, mid-thigh and the knee marker. The accurate palpation and placement of ASIS markers can cause less error in pelvic angles. The anterior/ posterior misplacement of thigh marker could cause a rotation of the knee and influence the Knee joint angles in the sagittal plane. The perpendicular distance of tibia and thigh markers to the line connecting posterior superior iliac spine -heel markers was estimated. The right thigh and tibia markers were attached 9.48 mm and 3.61 mm posteriorly in Lab 1. Backward misplacement of the thigh marker caused an external hip rotation offset and knee valgus in swing phase.

Another potential source of variably can arise from inconsistent anthropometric measures. However, Mean angles and S.D. values were still the same in both measurements.

Table 1: Standard deviations (deg)

	Lab 1	Lab 2	Lab1*
Pelvic tilt	1.99	1.59	1.99
Pelvic obliquity	0.77	0.89	0.77
Pelvic rotation	2.38	2.49	2.38
Hip flex/ext	3.52	2.73	3.54
Hip ab/adduction	1.50	1.57	1.51
Hip rotation	2.35	2.12	2.32
Knee flex/ext	4.77	3.48	4.79
Knee varus/valgus	1.91	1.15	1.74
Knee rotation	2.56	1.64	2.42
Ankle dorsi/plantar	3.47	2.35	3.44

*Lab1 with Lab 2 anthropometric measurements

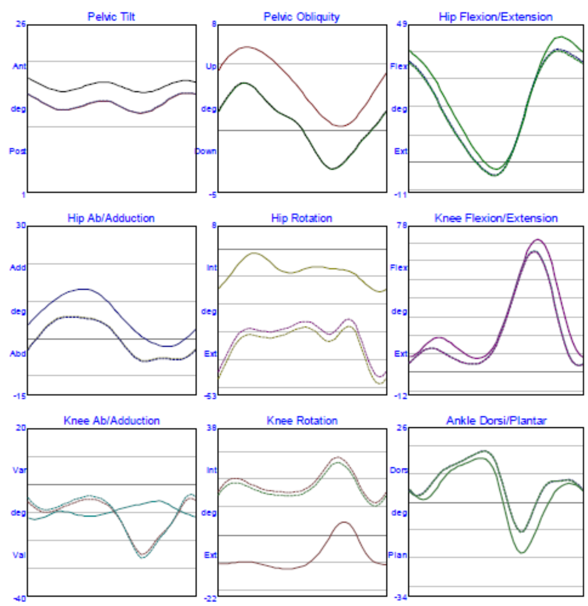

Figure 1: Average angle curves for Lab 1 (dashed lines) and Lab 2(Solid lines)

REFERENCES

1. Gorton, GE. Et al. (2009) Gait & posture, 29: 398-402
2. McGinley, JL., et al. (2009) Gait & Posture 29(3): 360-369.
3. Stolze, H. et al. (1998) Gait Posture, 7: 125-30

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DISCLOSURE STATEMENT

The authors have no conflicts of interest to disclose.

THE PROBLEM OF MULTIPLE COMPARISONS BETWEEN GROUPS OF TIME DEPENDENT DATA

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INTRODUCTION

One commonly encountered problem in the statistical analysis of time series is the need to determine which portions of the series are different between groups. Whereas it is standard practice to pick key points from curves and run the venerable analysis of variance (ANOVA) to determine if there is a difference between groups, this approach disregards information which may be valuable to the researcher. The advent of the functional ANOVA (FANOVA) [1] enabled researchers to effectively run comparisons between groups at all points within the time series. Critics of this method note that with the multiple comparisons required by time series, there is a need for alpha level correction which can lead to inflation of type II error. In recent work, Pataky et al. [2] proposed to solve this using random field theory (RFT) in combination with statistical parametric mapping (SPM) [3] to compare time series data in a point-wise manner.

Here, a method of functional comparison based on the commonly implemented LOESS to model the data in lieu of RFT is introduced. Statistical inference is then conducted using point-wise application of the Welch's *t* test in a manner similar to Pataky et al. [2], but with an adjusted Bonferroni correction that accounts for correlations in the *t* tests. Its usage is then demonstrated on time series data collected from foot pressure measurements and medial gastrocnemius forces from a public data set [4] that was previously analyzed using RFT [2].

CLINICAL SIGNIFICANCE

This methodology provides a means to statistically analyze trends between time series without cherry-picking points of interest and discarding potentially useful data.

METHODS

The first data set was comprised of longitudinally collected foot pressure data from typically developing (TD) children aged 2-14 years, and age matched children with cerebral palsy with GMFCS levels I and II (G12). Data were collected as part of an IRB approved study examining the longitudinal change in foot pressure in children. Only the lateral-mid-foot pressure (LMFP) data from this study are used here. These data were normalized to total pressure across the foot so that values could range from zero to 100%.

The second set of data are publicly available medial gastrocnemius forces estimated by an EMG-driven knee model and collected from individuals with knee pain during walking (N = 27), and individuals without pain (N = 16) [4]. These data were also adapted by Pataky et al. [2] in their investigation of using RFT to statistically compare time series between groups.

Data were modeled within each group using local polynomial regression fitting with LOESS in R. Modeled mean and standard deviation curves were sampled at equal time intervals to obtain point-wise estimates for these values at 4 sample rates. In longitudinal data the number of data points per group at each sample along the time domain were estimated using kernel density smoothing with endpoint adjustment. The resultant smoothed histogram was subsetted and then resampled at the same rate as for the mean and standard deviation curves to provide estimates of group sizes at each point. Finally, a point-wise Welch's two-sample *t* test was computed using these values. A new Bonferroni-based alpha-level correction was introduced to correct for correlation (ρ) in adjacent *P* values: $\alpha' = 0.05/[\ N(1-\rho^2) + \rho^2\ ]$.

DEMONSTRATION

Resampling data at different rates does not effectively change P values across the time series when LOESS is used. Likewise, by accounting for correlations between adjacent P values, it is possible to have a much less conservative alpha level corrections although these vary dependent on the number of comparisons. Unlike Bonferroni corrections, increasing the number of comparisons increases the alpha level because there are higher adjacent correlation coefficients. Data by group are presented in the left-hand panels of Fig 1, and log(p-values) are presented at the right. Alpha levels from top to bottom ($\alpha_{0.05}$, α'_{max}, α'_{min}, α_{Bonf}) are indicated by horizontal lines.

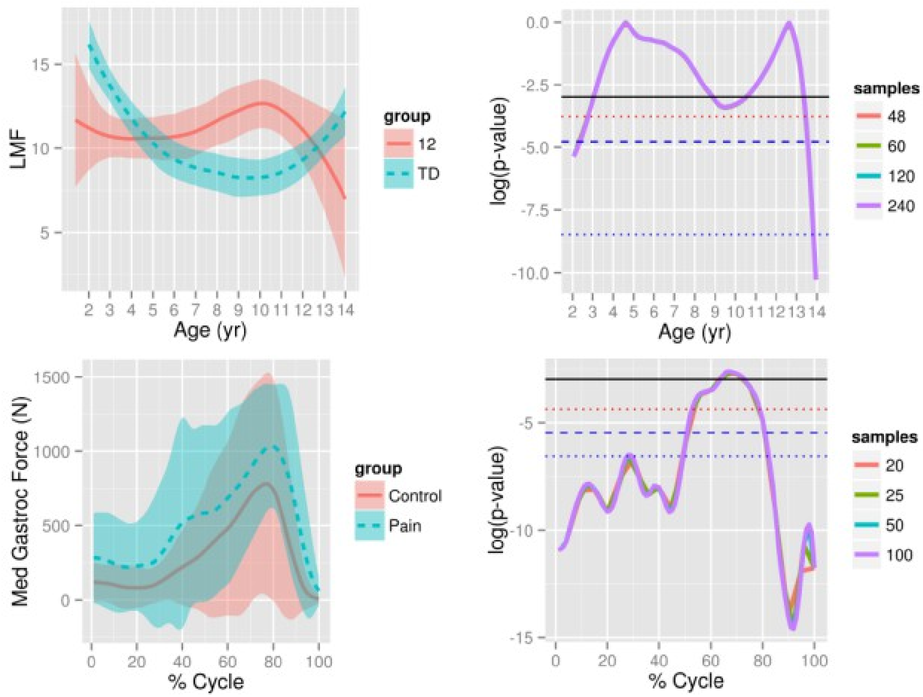

Figure 1

SUMMARY

Whereas it is common to focus on certain points in the motion (e.g., peak knee flexion) for statistical comparison between groups, this study highlighted the advantages of point-wise comparisons along the entire curves being compared. Although such methods have been suggested in the past, they were largely not used due to concerns about alpha level corrections that would result in unreachable significance levels. This study addressed these concerns through the use of LOESS to create a smoothed representation of the data and its variability which effectively correlates points that are adjacent in time. With such correlation it becomes extremely improbable that a type I error will occur anywhere across the time series. The tools utilized in conducting this type of analysis are readily available in most statistical packages and provide nearly identical results with more complex methodologies such as RFT, implying that this method may be valuable for analysis of motion capture data in the future.

REFERENCES

1. Ramsay, JO. (1982) Psychometrika, 47: 379-396
2. Pataky, TC, et al. (2015) J Biomech, 48: 186-189.
3. Friston, KJ, et al. (1994) Hum Brain Mapp. 1: 210-220.
4. Besier, TF, et al. (2009) J Biomech. 42:898-905.

DISCLOSURE STATEMENT

Timothy Niiler has no conflicts of interest to disclose.

Upper Extremity Biomechanical Analysis to Assess Thoracic and Pelvic Compensatory Strategies in Brachial Plexus Palsy Following Surgery

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INTRODUCTION

Brachial plexus palsy (BPP) is a debilitating condition resulting in paralysis and/or paresis to the affected limb. Incomplete BPP recovery typically results in an internally rotated and adducted shoulder position, requiring increased compensatory scapulothoracic motion [1]. Therapy targets the prevention of joint contractures, strengthening of muscles, and achievement of developmental milestones [1]. In the event that surgical intervention is required, the primary goal is to improve function of the affected limb; however, complete recovery is not expected, as asymmetries often remain identifiable [2]. Previous quantitative studies of BPP have established the abnormal and asymmetric kinematics present [3, 4], but assessments of post-surgical change are limited. Quantitative methods, using motion analysis, may prove useful in identifying and monitoring functional changes resulting from surgical intervention and rehabilitation. Our group has previously studied post-operative changes in movement symmetry in children with BPP [5]. The purpose of this study was to measure compensatory motions of the thorax and pelvis via 3D upper extremity motion analysis to quantify change following surgery in patients with BPP, using the Mallet assessment as a functional protocol.

CLINICAL SIGNIFICANCE

Compensatory motions of the pelvis and thorax may play a role in task performance in patients with BPP. Understanding these motions and how they change following surgery may help direct surgical planning and rehabilitation protocols in this population.

METHODS

This is a pilot study involving two children with BPP (10y F, 12y M). Both children were scheduled for arthroscopic anterior shoulder release. Each participant made an initial visit to the Motion Analysis Lab (MAL) within one day prior to surgery (Pre). Participants also completed a second visit 5-7 months post-surgery (Post). During each visit to the MAL, each subject was instrumented with reflective markers on the torso and upper extremity, following the model defined by Schnorenberg et al. [6]. Upper extremity kinematics were assessed while the subject performed the tasks of the Mallet assessment [7, 8]. Marker motion was captured using a 12-camera Raptor system (Motion Analysis Corp.; Santa Rosa, CA). Pelvic data were post-processed in Visual3D (C-Motion, Inc; Germantown, MD) and thoracic data were post-processed via a previously validated Matlab-based model (MathWorks, Inc.; Natick, MA) to calculate 3D kinematics [6].

Pelvic and thoracic movement compensations were measured in each plane for *Pre* and *Post* sessions by computing the angle-time integral from task initiation to termination (trapezoidal integration of the area between the kinematic trace and the resting position). Post-operative change was quantified as the difference between *Pre* and *Post*, with negative values indicating reduced compensation following surgery.

RESULTS

Sample data are provided in Figure 1. Both subjects demonstrate some degree of increased compensation following surgery. For the most part these increases in compensation did not align with any particular segment or plane. However, the shoulder abduction and hand-to-mouth tasks stood out for eliciting generalized increased compensation across both subjects in most segments and planes.

Some subject-specific effects did seem to be present, as Subject 2 demonstrated increased compensatory strategies for all tasks in nearly all segments and planes (Table 1). Changes in Subject 1 were more varied.

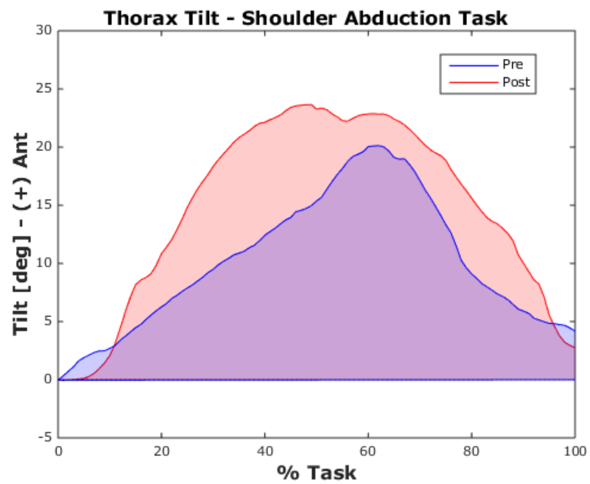

Figure 1: *Thorax tilt during the Mallet shoulder abduction task for Subject 2. Shaded areas represent compensation (deviation from resting position).*

Table 1: *Post-operative changes in pelvis and thorax strategies between pre and post sessions for Subject 2, measured as difference between kinematic integrals [deg-% task]. Positive values (gray shading) represent greater compensation post-surgery.*

Mallet Tasks	**Pelvis [deg-% task]**			**Thorax [deg-% task]**		
	Sag	*Cor*	*Tra*	*Sag*	*Cor*	*Tra*
External Rotation	87.8	54.7	0.4	702.2	81.1	134.7
Hand to Mouth	-32.4	284.7	169.1	123.1	106.8	39.4
Hand to Neck	-27.8	1.3	18.5	618.0	303.3	327.5
Hand to Stomach	107.3	149.5	0.2	263.5	-27.6	71.4
Shoulder Abduction	64.6	8.0	13.7	469.6	36.6	9.8

DISCUSSION

A clearer understanding of pelvic and thoracic compensatory strategies in patients before and after surgery for BPP may aid in surgical and rehabilitation planning. We expected to find decreased compensation post-surgery, but instead found the opposite. Of note, tasks involving shoulder abduction seemed to elicit greater motion of the pelvis and thorax from their starting positions. Subject-specific effects observed are potentially linked to the surgery performed; Subject 2 had a latissimus dorsi transfer performed via an open procedure, while all procedures performed on Subject 1 were done arthroscopically. Further study in a larger patient population is warranted to better understand these post-operative changes, and to study patterns in more distal joints.

REFERENCES

1. Waters (2005) J Pediatr Orthop, 25:116-126.
2. Hale et al. (2010) J Hand Surg Am, 35:322-332.
3. Duff et al. (2007) Clin Biomech, 22:630-638.
4. Russo et al. (2014) J Shoulder Elbow Surg, 23:327-338.
5. Garcia et al. (2016) 2016 Annual ASB Conference
6. Schnorenberg et al. (2014) J Biomech, 47:269-276.
7. Mallet (1972) Rev Chir Orthop Reparatrice Appar Mot, 58(1):166-168.

DISCLOSURE STATEMENT

The authors have no conflicts of interest to disclose.

Use of Motion Analysis and SmartWheel to Assist in Wheelchair Prescription: A Case Study

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INTRODUCTION

This case study was undertaken to evaluate the wheelchair (w/c) propulsion mechanics of an individual with myelomeningocele to assist with w/c prescription. Poor wheelchair propulsion mechanics and suboptimal wheelchair positioning have been linked to upper extremity injury and overuse. Analysis of push rim forces and kinematic variables can be valuable in w/c prescription, prevention of upper extremity injury/overuse and treatment for individuals with shoulder, elbow or wrist pain.

CLINICAL SIGNIFICANCE

The incidence of upper extremity pain is high in w/c users. It is imperative that clinicians provide the most efficient and comfortable mobility device for each individual w/c user as the cost of a manual w/c is very high and a replacement will usually only be funded after five years.

METHODS

Subject was an 18yo female with mid-lumbar level myelomeningocele who presented to spina bifida clinic with complaints of bilateral wrist and hand pain when propelling her w/c. She wanted to attend college in the fall and needed a new w/c to replace her older, heavier w/c. She was seen in the motion analysis laboratory (Time point 1 – T1) for an assessment of her propulsion abilities in her old wheelchair. Using a pair of 24" SmartWheels with pressure sensitive handrims on a standard tile floor, we were able to analyze push forces, frequency, length, arc of motion, smoothness as well as speed. The video motion capture allowed us to look at her w/c seating alignment. The subject returned (Time point 2 – T2) 16 months later after getting a new ultra-light w/c for re-evaluation.

RESULTS

Data was collected over four trials, averaged and compared to a current database of adult long term wheelchair users (norms). This subject's velocity, push frequency, push force normalized to body weight, push length or contact time on the rim, and push mechanical effectiveness (a measure of efficiency) were suboptimal (Table 1). Video review demonstrated that while she had a fairly good propulsion pattern at T1, she was sitting too high on her pressure relief cushion, her shoulders were elevated, the back canes interfered with her ability to have a longer arc of motion resulting in an increase in force delivery through the wheel rim over a shortened amount of contact time, and rapid movement into wrist extension at the end of push off. At T2 with her new w/c, she no longer had any wrist/pain and was able to travel to and from college on public transit and around campus without any issues. She was noted on video have changed her stroke pattern a bit and was now using excessive shoulder extension range with shoulder hiking prior to hitting the rim with her hands but was no longer snapping her wrists. At T2, she demonstrated improvements in speed, push frequency and push force, but had mildly decreased push length and push efficiency compared to T1. She was able to generate more force and propel faster with her new w/c but she would benefit from further education and training in proper propulsion techniques to enhance her overall efficiency.

Table 1. Propulsion parameters

Parameter	T1 (Old w/c)	T2 (New w/c)	Database average*
Speed (meters/sec)	0.90	1.10	1.29
Push frequency (contact/sec)	0.65	0.75	0.91
Push length (degrees)	51.3°	50.8°	74.5°
Push force - weight normalized (%)	4.8%	6.7%	11.3%

*Database averages shown are from user data collected by the SmartWheel User Group Database, and is for informational purposes only.

DISCUSSION

Overall analysis of the SmartWheel data at T1 showed that push force was low, the length of time the hand was on the rim was short and the push speed and mechanical effectiveness was low compared to our database of norms for this young adult. Her seat height and back rest and back cane heights were less than ideal. All of these findings combined may have contributed to her wrist and hand pain. She was instructed in optimal push mechanics using an elliptical pattern starting further back on the wheel using more force and less wrist extension at the end of each push. She was also given upper extremity stretching and strengthening exercises and modifications of wrist positioning during transfers and pressure reliefs. The results of the SmartWheel report were used during the new w/c evaluation completed by a physical therapist and w/c vendor. Based on the results, a larger ultra-light w/c was chosen with 24" wheels and minor seating adaptations from her previous w/c. With the new ultra-light w/c configuration, her wrist/hand pain was resolved and she was working on integrating improved push mechanics which increased her speed, push frequency and push force. She was able to generate more force and propel faster with her new w/c but she would benefit from further education and training in proper propulsion techniques to enhance her overall efficiency and technique. For this young lady, SmartWheel analysis was a valuable tool to assess push force, mechanics, speed and alignment in her old w/c which helped decision making for her new w/c as well as future training needs to prevent upper extremity pain in the future.

ACKNOWLEDGMENTS

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DISCLOSURE STATEMENT

None of the authors have conflicts of interest to disclose.