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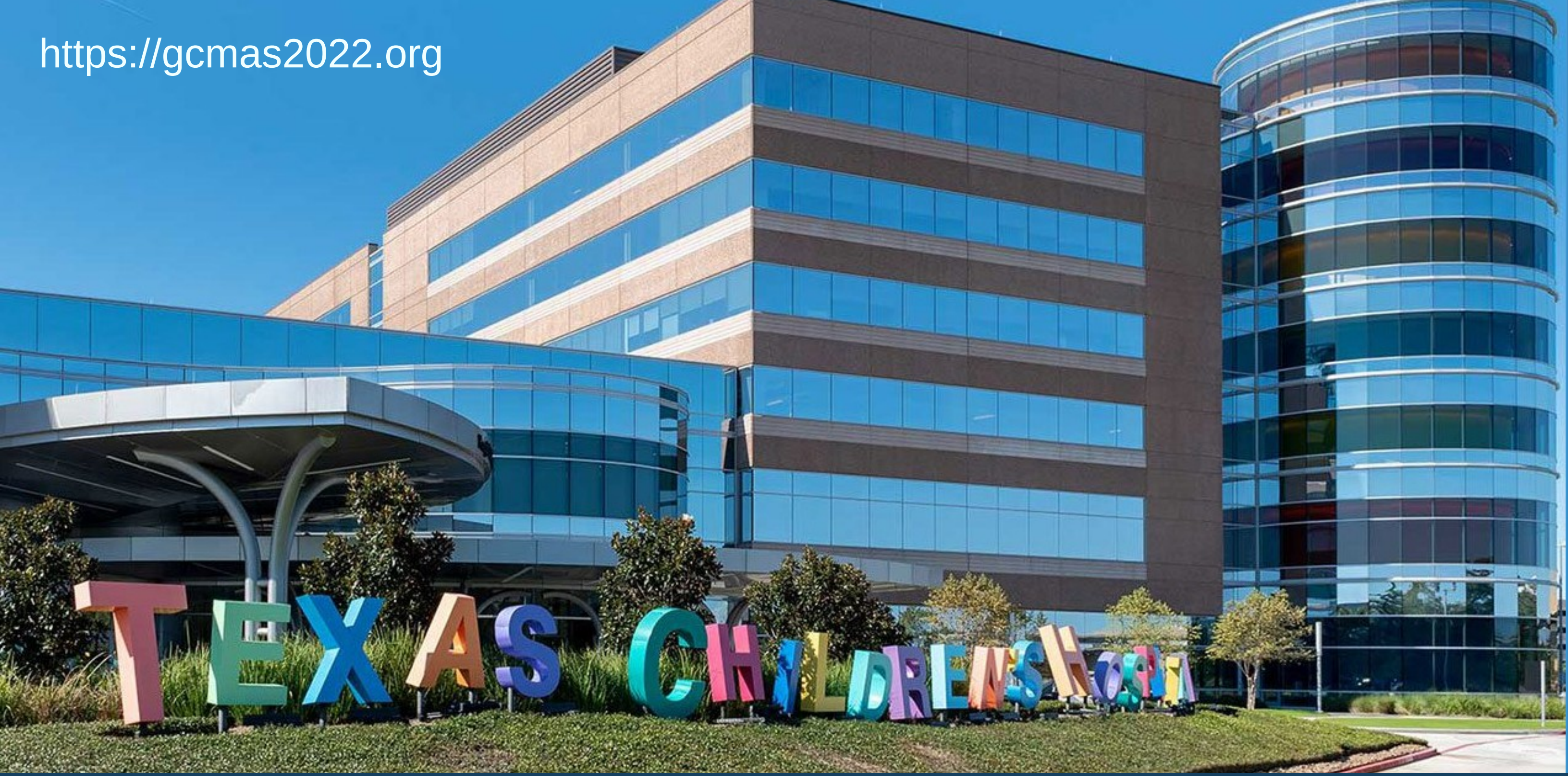

Abstracts of the 27th Annual (Virtual) Meeting

of the Gait & Clinical Movement Analysis Society

June 7-8, 2022

Improving functional outcomes of life through movement analysis since 1995

Abstracts of the 27th Annual (Virtual) Meeting

of the

Gait & Clinical Movement Analysis Society

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Schedule

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Keynote 1

Orthopedics

Poster Session 2

Data Analysis 1

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Tutorial 2: *5:30pm-6:30pm* Comparing Markerless And Marker-Based Motion Capture Systems For Clinical Motion Analysis Of Children And Youth With Atypical	**Poster 3**: *5:30pm-6:00pm* Poster 3	**Closing**: *5:30pm-6:00pm*	

Dear GCMAS members and supporters,

My name is Eric Dugan from Children's Hospital in Texas and it is my pleasure to welcome you to this year's annual GCMAS conference.

We were hoping to be able to have this meeting in person - I know that many of us are getting burned out of Zoom conferences, and we miss the in person interactions of typical conferences. However, with that being said, I think that we have a great conference lined up for you this year which I hope you will enjoy.

The foundation of the conference, as always, is the scientific program. We have over 75 presentations lined up over the next two days, and I'm really looking forward to hearing about the research being carried out across our community.

We are trying something a little different this year to facilitate participation at the conference. One of the drawbacks to a virtual conference is that we are probablycompeting for your attention to some degree. I suspect that some of you are going to be multitasking over the next two days trying to balance the conference with the many professional and personal responsibilities that you have. Now while these presentations will be posted and can be viewed later, I would highly encourage you to attend them realtime if you are able to so that you can engage with the presenters about their research.

In addition to the scientific program, I am really excited about our keynote speakers this year. I am really looking forward to Dr. Grabowski's and Dr. Steele's talks. It's a great opportunity to hear from two researchers at the top of their respective fields.

Dr. Grabowski will be speaking today at 1pm about whether athletes using running specific leg prostheses should be allowed to compete in the Olympic Games. Tomorrow, Dr. Steele will be speaking at one o'clock about motor control and understanding the role of altered motor control on cerebral palsy. I hope that you will be able to attend both of these talks. I think that they will be well worth your time.

We also have four great tutorials and special sessions lined up over the next two days. Each day we have a traditional tutorial and a special session on a topic relevant to clinical gait analysis laboratories. These tutorials are great for those attendees who may be relatively new to the field as well as those who have been doing this for a while. The special sessions are an opportunity to learn how other laboratories are tackling the issue of quality assurance and the introduction of new markerless motion capture technologies. I hope you find these sessions helpful and insightful as you move forward with your clinical and research opportunities.

And last, but not least, when organizing a conference like this you quickly realize how many people it takes to pull things off, and I need to say a lot of thank-yous. I would like to thank the GCMAS Board Members for their help and guidance through this process. I would also like to thank all of you who served as reviewers, and those who will be serving as moderators over the next two days. Your service is invaluable. I would also like to thank our keynote speakers for taking the time out of their busy schedules to share their work with us. I also need to thank Amy Barbuto and Karen Masterson who served as program co-chairs and helped with a lot of the behind the scenes work that needed to be done.

And finally, Tim Niiler who has been our website and technical guru. Tim has spent countless hours ensuring that we were able to create the platform necessary to hold this virtual conference.

Thank you for joining us this year, and I hope you have an amazing time and enjoy the conference.

Sincerely,
Eric Dugan
GCMAS2022 Conference Chair

Understanding Kinematic Relationships Within and Across Planes of Motion

Instructors: Sylvia Õunpuu, Kristan Pierz, Jennifer Rodriguez-MacClintic

Purpose: The purpose of this course is to demonstrate the role of motion analysis in gaining understanding of the relationship of joint and segment kinematics within and across planes of motion for a variety of gait pathologies.

Intended Audience: All personnel who are involved in the interpretation for treatment decision-making as well as the collection and processing of joint kinematic data. This includes, but is not limited to, physicians, physical therapists, kinesiologists, biomechanists and engineers.

Prerequisite Knowledge: Participants should have the basic understanding of how clinical gait analysis supports treatment decision-making, familiarity with gait analysis output and terminology related to kinematic data.

Outline of Course Content:

Clinical gait analysis data interpretation is a complex skill and requires general understanding of the gait pathology being studied as well as the specific joint level impairments for the patient being assessed. It also requires an understanding of the methods used to collect the motion data including the skeletal model that establishes the kinematic angle definitions. With this background knowledge, clinical gait analysis data can be interpreted for treatment decision-making purposes. During gait data interpretation, the primary goal is to define when gait pathology is present as a direct result of joint level impairments or muscle function and therefore a primary deviation. Gat deviations can also be secondary (at an adjacent joint) or a result of a voluntary compensation. The most appropriate treatment needs to be able to differentiate between these ways that a kinematic plot may not fall outside the reference data set.

This tutorial will have two components: a) background and definitions and, b) case studies to provide examples of interactions across planes that help to discern the difference between primary and secondary gait deviations and compensations. We will begin with an example of several joint angle definitions to emphasize the importance of this level of understanding joint kinematic data. We will then describe the definitions of within and cross plane interactions followed by primary, secondary and compensatory gait deviations.

This background will be followed by focused case examples of interactions across planes which will cover the majority of the tutorial. Each case study will illustrate a particular example of kinematic relationships within and across planes. These will be discussed in the context of primary, secondary and compensatory kinematic deviations. Most case examples will begin with a video segment and relevant clinical exam data for the particular joint of interest. These examples of within and across plane interactions will include pre and post-surgical outcomes data as well as barefoot and brace data that help to establish interactions across planes and what gait deviations are primary or other. The pathologies will include cerebral palsy as well as a variety of other neuromuscular gait disorders.

Learner-based Course Objectives:

At the completion of the course, the participant should:

To understand the importance of angle and segment definitions

To understand the definition of within plane kinematic interactions

To understand the definition of cross plane kinematic interactions

To be familiar with common examples of within and across plane interactions for a variety of diagnoses of neuromuscular origin

Schedule:

Speaker 1: Sylvia Ounpuu (1 minute): Introductions and purpose

Speaker 1: Sylvia Ounpuu (9 minutes): Definitions of within and across plane interactions

Speaker 1: Sylvia Ounpuu (10 minutes): Case examples – within plane

Speaker 2 & 3: Kristan Pierz and Jennifer Rodriguez-MacClintic (30 minutes): Case examples – across plane interactions (participants will be encouraged to stand if able to assist in understanding concepts)

Questions and answers (10 minutes)

IMPACT OF NEUROCOGNITIVE LOADING AND LIMB DOMINANCE ON HIP AND KNEE KINEMATICS IN A SINGLE-LEG CROSSOVER HOP

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INTRODUCTION

Anterior cruciate ligament (ACL) tears are well-studied injuries with significant repercussions for athletes. At the collegiate level, preseason physical performance tests (PPTs), are used to screen athletes for risk factors, including for ACL injuries.[1] Prospective research has established dynamic valgus and knee abduction moments as risk factors for ACL injuries.[2] Hop tests are commonly used PPTs, but incorporate only pre-planned movements and lack reactive decision-making components. Research suggests that anticipation has a large effect on knee mechanics associated with ACL injury, such that unplanned movements cause athletes to put themselves at greater risk of injury.[3] Thus, traditional PPTs may fail to properly identify athletes who are at risk for injury. To address this issue, Millikan et al. added a neurocognitive load to four commonly used PPTs, including a single-leg crossover hop, and found that healthy, moderately active participants exhibited decreases in performance when subject to the added neurocognitive load.[4,5] This study aimed to determine the impact of a neurocognitive load on lower limb biomechanics during a crossover hop test in a population of collegiate athletes.

CLINICAL SIGNIFICANCE

The addition of a neurocognitive load to a single leg crossover hop test could create a PPT that could more effectively screen athletes for injury risk and performance deficits.

METHODS

Twelve Division I athletes (11 females and 1 male) were recruited for this study. Using a 10-camera motion analysis system, sampling at 240 Hz, the locations of retro-reflective markers placed over specific bony landmarks on the subjects' head, trunk, pelvis, and bilateral upper and lower extremities were recorded during a hopping protocol. The protocol consisted of performing 16 trials of four consecutive single-leg crossover hops with and without a neurocognitive load. The neurocognitive load consisted of two lights in the athlete's peripheral vision at 70% of their height, 60cm in front of a start line. In a random sequence, one peripheral light would illuminate green, indicating the athlete was to initiate the hop sequence to that direction, and the other would illuminate red. Two non-hop trials (both lights illuminating blue) were randomly interspersed between the jump trials. The subjects completed 8 successful non-loaded trials on each extremity (no lights), 4 trials initially hopping to the right and 4 trials initially hopping to left on each extremity. Following these trials, the subjects repeated the sequences in the presence of the cognitive load. The marker coordinate data was filtered using a 4^{th} order, low pass Butterworth filter. For each lower extremity, the filtered data along with a 13 segment, 6 degree of freedom kinematic model were used to obtain the 3D angular displacements of the hip and knee joints at maximum knee valgus during the landing phase and the distance of the initial jump.

RESULTS

The effect of cognitive load and limb dominance on the sagittal and frontal plane position of the hip and knee joints are noted in the Table. With the exception of the effect of limb dominance on the sagittal plane position of the knee ($p = 0.01$), neither cognitive load nor limb dominance had a significant effect ($p > 0.05$) on joint position. Cognitive load and limb dominance had significant effects on the initial jump distance ($p < 0.001$ and $p = 0.01$, respectively). The initial jump distance was 127.5 (± 4) and 118.9 (± 4) mm and 137.4 (± 4) and 127.7 (± 2) mm for the dominant and non-dominant limbs during the cognitive and non-cognitive load trials, respectively.

Table. Impact of cognitive load and limb dominance on the mean (± 1 SEM) hip and knee joint kinematics at maximum knee valgus.

		Cognitive Load		Non cognitive load	
		Dominant	Non-Dominant	Dominant	Non-Dominant
Hip Joint	Sagittal Plane	27° ± 4	31° ± 4	31° ± 4	33° ± 4
	Frontal Plane	2° ± 2	-3° ± 2	2° ± 2	-4° ± 2
Knee Joint	Sagittal Plane	47° ± 3*	39° ± 3*	50° ± 3*	39° ± 3*
	Frontal Plane	6° ± 2	4° ± 2	6° ± 2	4° ± 2

* - significant effect of limb dominance p = 0.01

DISCUSSION

Our findings aligned with those of Millikan et al. Notably, cognitive load had significant effects on initial hop distance, suggesting the addition of a neurocognitive load creates a more challenging task. Given our small sample size, the lack of a significant effect of cognitive load on hip and knee joint position should be viewed with some caution. Additionally, only the joint position data at one time point was analyzed. Potentially the load may have impacted other kinetic or kinematic data (e.g., center of mass position). Further analysis of the effect of the load may be warranted. Future research should focus on the effects of a neurocognitive load on additional kinematic parameters using a larger sample size.

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Disclosure Statement

There are no conflicts of interest to disclose.

Relationships between hip strength and pitching biomechanics in adolescent baseball pitchers

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INTRODUCTION

Pitching is a dynamic task which creates motions that are stressful on the upper extremity, reaching levels of forces up to 5 times a pitcher's body weight [1]. Throughout the course of a pitchers career the repetitive throwing motion increases the risk for injury at the shoulder and elbow joints, which has already been well-recognized as a problem in baseball pitchers [2]. The pitching cycle has been described as a kinetic chain deriving energy from the lower extremities, and transferring this energy via pelvic and trunk rotation to the upper extremities to produce throwing velocity [3]. It has been hypothesized that strengthening of the lower extremities, an origin of the energy transitioned into the throwing arm, may help enhance performance as well as avoid injury. The purpose of this study was to identify associations between hip strength and pitching biomechanics in adolescent baseball pitchers.

CLINICAL SIGNIFICANCE

Understanding how hip strength contributes to pitching biomechanics with the use of 3D motion analysis on individual body segments can lead to a better explanation for the creation of excessive forces on the shoulder and elbow joints. This may then be translated to coaches and athletic trainers for injury prevention.

METHODS

26 right-handed adolescent male baseball pitchers participated (16.1 ± 0.8 years, 77.5 ± 8.5 kg, 184.3 ± 5.5 cm). Each pitcher had at least 3 years of pitching experience with no current arm pain or history of throwing arm surgery. A single certified physical therapist collected all strength measurements using a handheld dynamometer. 8 cameras (Motion Analysis Corporation, Santa Rosa, CA) were positioned around an artificial mound to collect data at 300 Hz. 47 reflective markers were attached to the subject and 10 fastball pitches were recorded. Marker data was identified and filtered in Cortex software, and kinematics and kinetics were calculated using Visual 3D software (C-Motion, Germantown, MD). 13 kinematic metrics were analyzed at foot contact (FC), maximum shoulder external rotation (MER), and ball release (BR). Kinetic metrics analyzed included peak elbow varus torque (EVT) and peak shoulder internal rotation torque (SIRT). Torques were normalized by body mass (BM) and height (H). Statistical analysis was completed using SPSS with a significance level of $p < 0.01$ used.

RESULTS

10 correlations were statistically significant at a p-value < 0.01 and a $|r| > .5$ (Table 1), demonstrating strong correlations with hip strength. At MER, shoulder horizontal abduction angle and lead hip internal rotation strength, shoulder horizontal abduction angle and back hip internal rotation strength, shoulder horizontal abduction angle and back hip external rotation

strength, and hip to shoulder separation angle and back hip extension strength were all found to be significantly correlated. At BR, shoulder horizontal abduction angle and lead hip internal rotation strength, shoulder horizontal abduction angle and back hip internal rotation strength, hip to shoulder separation angle and back hip extension strength, and back hip adduction angle and lead hip internal rotation strength were found to be significantly correlated. When analyzing the kinetics, normalized elbow varus torque was strongly correlated with both lead hip abduction strength and back hip abduction strength.

Table 1: Significant correlations between pitching biomechanics and hip strength.

Kinematics at MER	r	p-value
Shoulder horizontal abduction angle and lead hip internal rotation strength	-0.564	0.003
Shoulder horizontal abduction angle and back hip internal rotation strength	-0.528	0.006
Shoulder horizontal abduction angle and back hip external rotation strength	-0.501	0.009
Hip-to-shoulder separation angle and back hip extension strength	0.538	0.005
Kinematics at BR		
Shoulder horizontal abduction angle and lead hip internal rotation strength	-0.554	0.003
Shoulder horizontal abduction angle and back hip internal rotation strength	-0.520	0.006
Hip-to-shoulder separation angle and back hip extension strength	0.562	0.003
Back hip adduction angle and lead hip internal rotation strength	0.512	0.008
Kinetics		
Normalized elbow varus torque and lead hip abduction strength	0.619	0.001
Normalized elbow varus torque and back hip abduction strength	0.565	0.003

DISCUSSION

The results demonstrate strong correlations within the relationship between hip strength and pitching biomechanics. Specifically, elbow varus torque was correlated with both lead and back hip abduction strength. This may be linked to greater pelvic rotation, allowing the transfer of forces to the lead leg. EVT is an important measure when evaluating potential elbow injury and keeping the forces below joint threshold is key. Overall, our data supports the importance of hip strength throughout the throwing motion and encourages pitchers to include hip strengthening exercises within their workouts.

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ACKNOWLEDGMENTS

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DISCLOSURE STATEMENTS

The authors have nothing to disclose.

MODELING THE TRUNK FOR SPORTS TESTING IN A MOTION CAPTURE LAB: MARKER PLACEMENT, SEGMENT DEFINITION, AND TRACKING

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INTRODUCTION

Trunk position can rapidly change during dynamic, game-like movements such as jumping, landing, or cutting, influencing an athlete's center-of-mass and overall stability [1]. Thus, recent literature has emphasized the importance of evaluating trunk kinematics for return-to-play assessments in athletic populations [1]. Recent reports have shown that reduced trunk flexion and greater trunk lean over the stance limb have been identified as strong risk factors for knee injury [1]. This is especially concerning in female athletes given that they naturally exhibit increased knee valgus. However, while sex differences are commonly investigated in biomechanical studies [2], the clear anatomical differences between the two groups are not always considered when testing. Specifically, marker placement and modeling alternatives are essential to ensure trunk motion in female patients is accurately and reliably assessed.

When utilizing motion capture technology, the trunk is typically defined by four anatomical landmarks: base of both the cervical and thoracic spines, proximal and distal ends of the sternum. The purpose of the current study was to assess trunk model variations to determine a marker set that could define and track the trunk reliably across patients and motion capture labs.

CLINICAL SIGNIFICANCE

As motion capture technology becomes more commonly used for clinical decision-making, it is essential that biomechanical risk factors, such as trunk position, are accurately modeled and assessed in both male and female patients.

METHODS

A convenience sample of eighteen healthy participants (7 males; age: 21.8, SD 4.1 years; height: 168.6, SD 10.4 cm; weight: 68.2, SD 15.3 kg) were tested. For motion analysis testing, participants were instrumented with retroreflective markers placed on the following landmarks: jugular notch (CLAV), mid-sternum (M-STRN), low sternum (L-STRN), xiphoid process (XP), 7th cervical spinous process (C7), and the 1st, 4th, 8th, 10th, and 12th thoracic spinous processes (T1, T4, T8, T10, and T12, respectively). A 14-camera motion capture system (Vicon, Denver, CO, USA) was used to collect kinematic data, sampling at 240Hz while participants performed a standing trial, over-ground gait, and a drop vertical jump (DVJ). Intraclass correlation coefficients (ICCs) were computed to assess the agreement between trunk models. Lastly, static offsets were computed by comparing the average position of the recommended model to the alternative models in each plane.

DEMONSTRATION

Trunk kinematics were most influenced by variations in sternum marker placement, primarily in the sagittal plane during gait (poor to moderate, ICCs: 0.36-0.60). Less agreement in sagittal plane trunk kinematics was also due to both cervical (moderate to good, ICCs: 0.71-0.85) and thorax marker placement (moderate to good, ICCs: 0.54-0.79). Model differences in the coronal and transverse planes were primarily due to thorax marker placement across all tasks (good to excellent, coronal ICCs: 0.81-0.96, transverse ICCs: 0.84-0.98; DVJ Lean seen in Figure 1). Overall, the transverse plane was least influenced by marker placement. Lastly, static offsets between model variations remained less than 4 degrees across planes (Table 1) and were greatest in the sagittal plane.

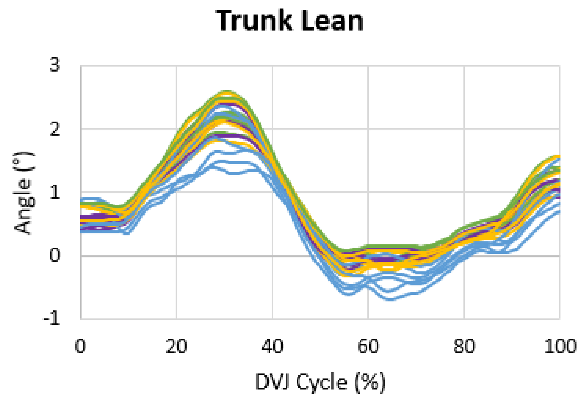

Figure 1: DVJ trunk lean for all trunk model variations. Note: **T4**, **T8**, **T10**, and **T12** models are highlighted in different colors.

Table 1: Offsets (degrees) for alternative models with individual marker substitutions compared to recommended model (CLAV, XP, T1, T10).

Alternate Models by Marker				Sagittal	Coronal	Transverse
CLAV	**MS**	T1	T10	-3.74	-0.16	0.25
CLAV	**LS**	T1	T10	0.91	-0.18	0.16
CLAV	XP	**C7**	T10	-1.99	-0.07	0.10
CLAV	XP	T1	**T4**	1.51	0.14	0.11
CLAV	XP	T1	**T8**	-0.62	0.05	0.06
CLAV	XP	T1	**T12**	1.07	-0.01	0.03

SUMMARY

The authors recommend defining the trunk segment using the CLAV, XP, T1, and T10 markers, and tracking the trunk during dynamic tasks with the M-STRN instead of the XP. If a different trunk marker set has been adopted, the offsets presented here may be used to adjust accordingly. All markers should be placed directly on the skin. If clothing for any patient proves problematic, a dypstick may be used during the static trial, and marker placement may be adjusted slightly for dynamic trials. For example, the T10 can be placed slightly more proximal towards the T8 placement but not lower towards T12. These recommendations will help ensure that trunk kinematics collected across patients and/or labs are consistent and reliable.

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DISCLOSURE STATEMENT Co-authors Kirsten Tulchin-Francis and Ross Chafetz are the current president and secretary-treasurer of GCMAS. No other conflicts to disclose.

DIFFERENCES IN THE NON-SAGITTAL KNEE MOMENT VECTOR-FIELD BETWEEN LAND AND CUT MANEUVERS AND DROP LANDINGS

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INTRODUCTION

Approximately 40% of the estimated 120,000+ anterior cruciate ligament (ACL) injuries reported yearly are the result of noncontact mechanisms [1,2]. These ACL injuries are the result of multi-joint, multi-planar mechanisms that occur within the first 100ms after initial contact during landing or cutting manoeuvres [3]. In a unique study of 10 ACL injuries in female handball and basketball players, it was determined that injury occurred at approximately 40ms after initial contact and that it corresponded to rapid knee valgus and internal tibial rotation [4]. Recently it has been reported that non-sagittal knee moments may identify individuals at risk for ACL injury better than uni-planar moments [5]. The multi-planar nature of ACL injury should be considered when assessing injury risk, selecting functional movement tasks, and developing criteria to guide return to play decisions. Therefore, the purpose of this study was to determine if there are between task differences in non-sagittal plane knee moments during a single leg land-and-cut maneuver and a single leg drop landing.

CLINICAL SIGNIFICANCE

Understanding the biomechanical demands of different types of landing tasks may help clinicians implement them at appropriate times relative to injury and clearance during rehabilitation following ACL injury.

METHODS

Forty-one recreationally active individuals (71.0±14.2 kg, 1.71±0.10 m) performed unilateral drop landings as well as land-and-cuts manoeuvres from a 40 cm box on their dominant legs. The dominant leg was defined as the preferred kicking leg. Prior to data collection, participants were informed of study procedures and provided written informed consent in accordance with institutional guidelines. Marker coordinate data (200 Hz, Vicon) and ground reaction force data (2000 Hz, Bertec) were collected simultaneously for all trials. Marker coordinate and GRF data were filtered using fourth-order Butterworth low-pass filters with cut-off frequency of 15 Hz. 3D joint kinematics were calculated using a joint coordinate system approach. 3D internal joint moments were calculated using standard inverse dynamics techniques and expressed in the joint coordinate system. Joint moments were normalized to body mass.

Statistical parametric mapping paired Hotelling's T^2 test was used to assess between task differences in knee adduction and knee rotation moment vector-field [7]. If significant differences were observed, post hoc SPM paired-sample t-test were performed on each vector component. Significance for all tests was *a priori* set at $p < 0.05$.

RESULTS

Hotelling's T^2 (SPM{T^2}) revealed significant differences in non-sagittal knee moment vector-field between 0-4ms; 18-54ms; and 58 -100ms after IC (Figure 1A). Follow-up post hoc t-tests (SPM{t}) revealed differences in My between 0-4ms; 18-55ms; and 72 -100ms after IC (Figure 1B) and differences in Mz between 0-4ms; 18-48ms; and 68 -100ms after IC (Figure 1C).

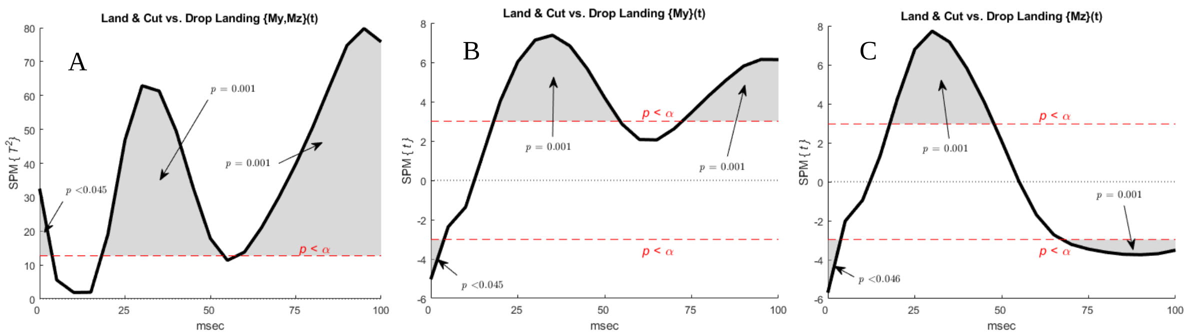

Figure 1. (A) Hotelling's paired T^2 trajectory (SPM{T^2}) representing differences in non-sagittal plane knee moments between land and cut and drop landing tasks. Post-hoc scalar t-tests (SPM{t}) on the **(B)** knee adduction moment (My) and **(C)** knee internal rotation moment (Mz).

DISCUSSION

Differences in non-sagittal plane knee moments between the two landing tasks were observed. Interestingly, the differences observed may not indicate that one task is safer than the other. Focusing on the differences that span the 40ms timeframe, the post-hoc analysis revealed that the land and cut task had higher internal knee adduction moments while the drop landing had larger external knee rotation moments. This is particularly relevant as this is the timeframe that increased non-sagittal plane motions and injury occurs [4,6]. Also of note, the differences in non-sagittal plane knee moments in the current study occur almost immediately after initial contact. It seems reasonable to expect differences in non-sagittal knee moments over the course of the entire landing stance phase of the movements due to the differing lateral and rotational requirements. However, the differences observed in the current study occur prior to these motions. The differences in non-sagittal plane knee moments in the current study may be influenced by different pre-landing strategies in response to differing task goals. Future work should continue to explore these differences in the context of developing optimal assessment batteries as part rehabilitation programs following ACL injury.

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DISCLOSURE STATEMENT

Authors have no disclosures to make.

Biomechanical differences between consecutive landings of a drop vertical jump task in collegiate aged females

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INTRODUCTION

Female athletes are subject to a fourfold increased risk of non-contact anterior cruciate ligament (ACL) injury when compared to male athletes, most notably in sports requiring lateral cuts and jump landing maneuvers [1]. A double leg drop vertical jump (DVJ) task is commonly used in ACL injury risk prediction, with emphasis historically placed on the first landing of the DVJ. Recent evidence demonstrates the second landing of a DVJ to be more representative of ACL injury risk due to athletes displaying less neuromuscular control with unplanned landings, which more closely simulates in-game play [2]. This study aimed to investigate the kinetics and kinematics of the hip, knee, and ankle joints at the time of peak knee abduction moment (KAM) during consecutive DVJ landings. It was hypothesized that there would be biomechanical differences between the first and second landings.

CLINICAL SIGNIFICANCE

Due to the increased similarity to in-game mechanics in unplanned landings, analysis of the second landing can provide a more encompassing evaluation of an individual's risk when compared to the first landing. These results may be used to identify individuals most at risk for ACL tear so athletes can begin injury prevention programs.

METHODS

Twenty-one active college-aged females (21.0 ± 1.7 years, 63.7 ± 5.2 kg, 1.7 ± 0.1 m) were recruited to perform a double leg DVJ task following protocol approved by the Medical College of Wisconsin institutional review board. Seventeen reflective markers were placed on each participant following a modified Helen Hayes marker set used for the Plug-in Gait model (Vicon Motion Systems, Ltd.; Oxford, UK). A twelve-camera Vicon MX motion analysis system sampling at 250 Hz was used to collect 3D motion data while in-ground force plates sampling at 3000 Hz were used to measure ground reaction forces (Bertec Corporation, Columbus, OH). Each subject completed 10 successful double leg DVJs. These consisted of stepping off a 30 cm plyometric box, landing simultaneously on both legs on separate force plates, immediately completing one maximal height jump, and again landing on both legs simultaneously on individual force plates. Data was processed as previously described [3]. Paired-samples t-tests were conducted to compare the means of the kinetic and kinematic variables at peak KAM of the first and second landing.

RESULTS

Four metrics were statistically significant between the first and second landings (Table 1). The sagittal knee angle was significantly higher for the first landing than the second landing ($t_{20} = 4.138$, $p = 0.001$), the sagittal hip angle was significantly higher for the first landing than the second landing ($t_{20} = 4.977$, $p = 0.000$), the sagittal knee moment was significantly higher for the first landing than the second landing (t_{20} = 5.702, p = 0.000), and the anterior tibial shear force was significantly higher for the first landing than the second landing (t_{20} = 4.479, p = 0.000). All variables were calculated based on data from the left leg due to absence of statistically significant side-to-side asymmetries.

Table 1: Mean and standard deviations (SD) between landing 1 and landing 2.

	Landing 1			Landing 2			
	Mean	±	SD	Mean	±	SD	p-value
Base of support (mm)	116.2	±	53.5	128.2	±	41.9	0.119
Time of peak knee coronal moment (ms)	54.2	±	20.8	58.6	±	23.0	0.392
Foot progression angle (°)	-4.5	±	4.4	-2.8	±	7.5	0.334
Sagittal knee angle (°)	48.7	±	12.4	40.7	±	11.6	0.001*
Coronal knee angle (°)	6.1	±	8.2	5.7	±	6.8	0.566
Sagittal hip angle (°)	48.0	±	9.8	39.3	±	8.6	0.000*
Coronal hip angle (°)	-9.8	±	5.5	-10.7	±	5.1	0.106
Sagittal knee moment (Nm/kg)	1.49	±	0.50	0.88	±	0.56	0.000*
Coronal knee moment (Nm/kg)	0.75	±	0.41	0.60	±	0.40	0.125
Transverse knee moment (Nm/kg)	0.07	±	0.08	0.07	±	0.07	0.586
Sagittal hip moment (Nm/kg)	-1.19	±	1.53	-0.74	±	1.38	0.261
Anterior tibial shear force (N/kg)	6.78	±	2.22	4.57	±	2.28	0.000*

*Significant at $p < 0.05$

DISCUSSION

We found participants exhibited significantly different biomechanics between landings. Confirming previous findings, knee and hip angles in the sagittal plane were decreased during the second landing phase, with a greater decrease at the hip joint [2]. Decreased sagittal joint angles may be explained by absence of loading necessity for a subsequent jump post second landing. This erect landing places an individual in a more perilous kinematic position, resulting in higher risk of ACL injury. We also found decreased sagittal moment of the knee joint and decreased anterior tibial shear force during the second landing, both of which suggest lesser ACL strain at peak KAM. These results that differ from previous investigations may be explained by athletes gaining greater neuromuscular control with age, thus displaying less risky landing behaviors.

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ACKNOWLEDGMENTS

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DISCLOSURE STATEMENTS

The authors have nothing to disclose.

GROUND REACTION FORCE ASYMMETRY DURING SIMULATED SPORTS ACTIVITIES TO BETTER UNDERSTAND HEMOPHILIC ARTHROPATHY

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INTRODUCTION

Persons with hemophilia (PwH) have low clotting factor levels, leading to increased risk of bleeding into large joints, causing long-term joint damage known as hemophilic arthropathy. Prophylactic medications increase clotting factor levels and decrease but do not eliminate joint damage [1]. In clinical care, the degree of joint damage is routinely characterized using the Hemophilia Joint Health Score (HJHS), a validated physical exam score that assesses swelling, crepitus, pain, and range of motion, with higher scores indicating more severe arthropathy [2].

Increased and asymmetric joint loading is a known contributor to joint damage [3,4], but has not been rigorously investigated in PwH. We hypothesize that asymmetric movement patterns in individuals with clinical joint arthropathy could promote force asymmetry, which may lead to increased joint bleeding risk. Through an ongoing study, we are evaluating the relationship between hemophilic arthropathy, movement asymmetry measured with computerized motion analysis, and bleeding risk. Here we compare ground reaction forces (GRF) during sports tasks performed by 2 PwH with discordant HJHS scores to assess presence of force asymmetry.

CLINICAL SIGNIFICANCE

A better understanding of movement patterns related to hemophilic arthropathy will allow development of targeted physical therapy programs to prevent joint bleeding and pain.

METHODS

HJHS evaluations were performed by experienced physical therapists specifically trained in hemophilia care. To minimize bleeding risk during the study, participants took prophylactic medication as they would when participating in sports. Kinematic data were collected using 8 Vicon vantage cameras (120Hz) combined with synchronized GRF data from 6 Bertec force plates (960Hz). Participants performed barefoot walk, shoe walk, single leg (SL) vertical hop, SL horizontal hop, drop jump, 45° cut, deceleration, double leg (DL) squat, and SL squat with consistent cueing, wearing preferred athletic footwear. For drop jump, participants dropped off a 31cm box and immediately performed a maximum vertical jump. Deceleration consisted of a running approach with a 2-legged jump stop.

Analog force plate signals were filtered in Visual3D using a 4th order Butterworth filter with cutoff frequencies of 10 Hz (walks, squats) and 100 Hz (hops, cut, deceleration, and drop jump). To objectively evaluate symmetry, peak vertical GRF (vGRF) were compared for each activity using the Normalized Symmetry Index [5], with the *more affected leg* defined using the *greater* (worse) HJHS score. A negative NSI indicates the more affected leg demonstrated the larger peak vGRF compared to the contralateral leg, and an NSI of zero indicates absolute symmetry. The protocol was approved by the institutional review board (#19-3072).

RESULTS

Table 1 shows NSI values for peak vGRF for each activity, and HJHS evaluations. *Subject A* has minimal hemophilic arthropathy balanced between lower extremities, and *subject B* has more hemophilic arthropathy in the right leg (Table 1). Subject B (higher HJHS asymmetry) had more vGRF asymmetry than subject A in all tasks except SL squat. However, subject B had deeper

Table 1. NSI of peak vGRF during each activity and HJHS.											
BF = barefoot, SL = single leg, DL = double leg, DJ = drop jump, LE = lower extremity, BW = body weight											
		Less Affected (Better HJHS) LE Peak vGRF			More Affected (Worse HJHS) LE Peak vGRF				HJHS		
		n	mean (% bw)	stdev	n	mean (% bw)	stdev	Mean NSI		Left	Right
Participant A	BF Walk	9	117%	3%	6	118%	1%	-1.36%	Elbow	0	1 (+1)
	Shoe Walk	7	119%	3%	6	122%	4%	-2.63%	Knee	1	2
	SL Squat	9	126%	3%	8	123%	3%	3.30%	Ankle	4 (+2)	2 (-1)
	DL Squat	14	67%	2%	14	61%	2%	12.53%	LE Total	**5**	**4**
	SL Vert Hop	12	375%	31%	15	388%	54%	-2.86%	*higher HJHS = worse hemophilic arthropathy*		
	SL Horiz Hop	3	352%	11%	3	344%	9%	2.08%			
	Deceleration	6	287%	49%	6	277%	55%	2.65%			
	Cut	3	275%	56%	3	310%	52%	-9.44%			
	DJ Drop	3	268%	29%	3	315%	60%	-13.16%	*parentheses = change from previous exam*		
	DJ Jump	3	239%	115%	3	339%	138%	-20.47%			
Participant B	BF Walk	8	110%	2%	5	106%	3%	3.90%	Elbow	7	0 (-1)
	Shoe Walk	6	111%	7%	5	105%	5%	4.43%	Knee	1	9
	SL Squat	9	118%	3%	9	119%	2%	-2.11%	Ankle	1 (-2)	7
	DL Squat	14	100%	3%	14	68%	3%	41.18%	LE Total	**2**	**16**
	SL Vert Hop	9	278%	26%	7	252%	15%	7.63%			
	SL Horiz Hop	3	271%	36%	3	260%	22%	3.54%			
	Deceleration	6	250%	32%	6	203%	24%	15.04%			
	Cut	3	205%	10%	3	217%	27%	-4.81%			
	DJ Drop	3	161%	25%	3	134%	13%	14.20%			
	DJ Jump	3	315%	47%	3	145%	5%	46.77%			

squats in the less affected leg (17.9% of standing height vs 11.5% in more affected leg) while subject A's squat depths were similar (18.3%, 18.8%). Although horizontal hop vGRF were relatively symmetric for subject B, braking GRF was more asymmetric (mean NSI 24.7% for B, vs 2.08% for A), and horizontal hop height and distance (% standing ht) were more asymmetric for B (B: more affected ht/dist 3.1%/ 39.4% vs 7.4%/66.7%; A: more affected ht/dist 8.7%/89.6% vs 10.8%/96.2%). For both subjects, walking tasks had the best symmetry, followed by single-leg tasks, then tasks in which both legs acted simultaneously. Single leg and higher velocity tasks overall had higher peak vGRF. Subject B had negative NSI values (higher peak vGRF in more affected leg) in only 2 activities, while subject A favored the more affected leg in over half of activities.

DISCUSSION

PwH with more joint damage are at higher risk of future joint bleeding, for reasons incompletely understood. In this study, a PwH with more severe and asymmetric joint damage had more asymmetric force application between legs, and more frequently shifted weight onto the less affected leg, compared to a PwH with symmetric and less severe joint damage. Loading asymmetry could predispose PwH to joint bleeding, both because decreased loading of the more affected leg could lead to muscle weakness and diminished joint stability, and because compensation with the less affected leg creates excess loading through the joints. The ongoing study will collect bleeding event data in a total of 90 PwH to test this hypothesis. Asymmetry and peak forces differed between tasks, supporting the need for a battery of tests. Our eventual goal is to develop a hemophilia "stress test" to better characterize and prevent movement-related bleeding.

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THE IMPACT OF TIME ON ANKLE FUNCTION IN CHILDREN WITH CHARCOT-MARIE-TOOTH DISEASE

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INTRODUCTION

Charcot-Marie-Tooth disease (CMT) is a hereditary neuropathy that causes distal weakness with associated gait deviations of varying severity especially at the ankle. Additionally, CMT is progressive and can result in increasing weakness and related gait decline over time. It has been shown that gait velocity does not increase with age at the same rate as peers for children with CMT. However, it is not known how ankle function during gait may change. Therefore, the purpose of this study was to document longitudinal changes in ankle function over time in children with CMT Type 1 (CMT1) and 2 (CMT2).

CLINICAL SIGNIFICANCE

Understanding gait decline is critical for evaluating treatment outcomes including orthopedic surgery and potential new therapies that aim to reduce the impact of CMT on gait.

METHODS

Forty-five children with CMT (34 CMT1, 11 CMT2; 28 males; mean age 12.1, SD 4.1 years) underwent comprehensive gait analysis using the conventional gait model walking barefoot at a self-selected pace. Gait assessments included standard clinical examination (strength, range of motion (ROM), anthropometrics), kinematics and kinetics (with non-dimensional normalization) and temporal-spatial parameters. 17 of these patients had repeat gait assessments (range 2 to 4 tests) without intervening surgery. To evaluate the change over time in select ankle kinematic and kinetic variables, linear mixed effect models were used with age/visit number and side as fixed effects and a random intercept for participant to account for the repeated measures. The outcome variables included peak dorsiflexion (DF) in terminal stance (TST, 45-65% of gait cycle (GC)), peak DF in mid-swing (MSW, 70-90% GC) and peak internal ankle plantar flexor (PF) moment and power generation in TST.

RESULTS

Strength: Plantar flexor ($p=0.01$) and dorsiflexor ($p=0.046$) strength were lower in CMT2 compared with CMT1. However, there was no relationship between strength and age for either muscle group or CMT type. Passive ROM: There was no relationship between passive DF ROM and age for either CMT type. Peak DF TST: Patients with CMT1 generally had reduced peak DF in TST at a younger age, which increased through approximately age 13 years ($p=0.004$) and then plateaued in the typically developing (TD) range ($p=0.73$, Fig 1a). Patients with CMT2 generally had typical or increased peak DF in TST which did not change with age ($p=0.88$, Fig 1b). Peak DF TST was greater if PF strength <3 in CMT1 ($p<0.001$). PF strength did not relate to DF TST in CMT2. Peak DF MSW: In CMT1, peak DF MSW was closely related to peak DF TST ($p<0.001$) and followed the same pattern, increasing through age 13 years ($p=0.005$) and then plateauing ($p=0.99$, Fig 2a). In CMT2, peak ankle DF in MSW was not related to peak DF in TST ($p=0.41$) or age ($p=0.88$, Fig 2b). Peak DF MSW was lower if DF strength <3 in CMT1 or <4 in CMT2 ($p<0.02$). A few patients did not fit these kinematic patterns due to large plantar flexor contracture and/or significant weakness.

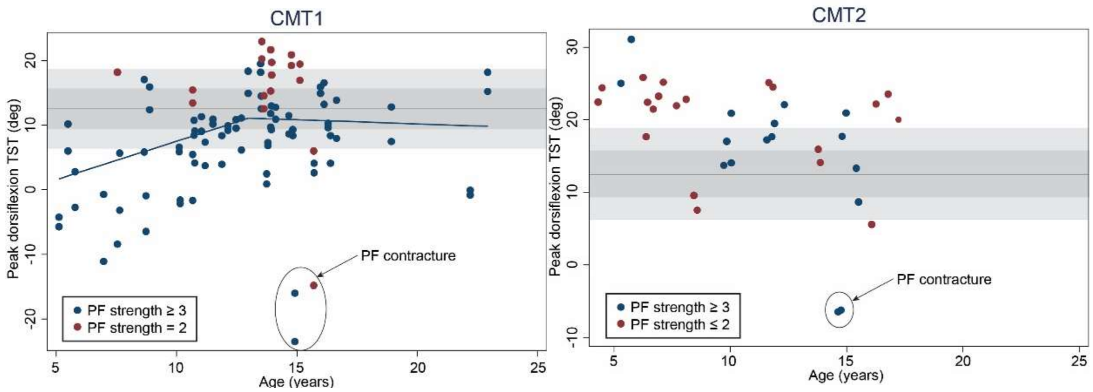

Figure 1: Peak dorsiflexion terminal stance vs. age a) CMT1 and b) CMT2.

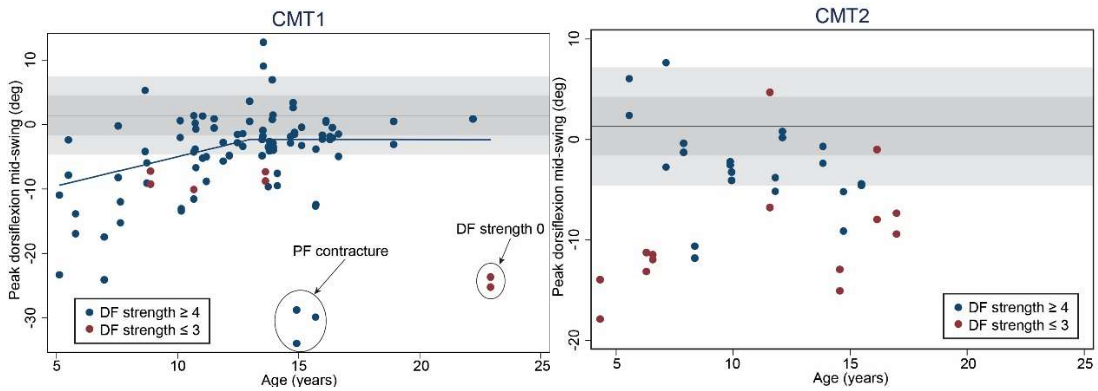

Figure 2: Peak dorsiflexion in mid-swing vs. age a) CMT1 and b) CMT2.

Peak PF moment in stance: There was no relationship between peak PF moment in stance and age for either CMT group ($p>0.18$). Peak PF power: In CMT1, peak PF power increased with age ($p=0.001$). There was no relationship between peak PF power and age in CMT2 ($p=0.49$).

DISCUSSION

Patients with CMT1 showed closely correlated patterns of increasing peak DF in TST and MSW to approximately 13 years of age, plateauing in the TD range. These results cannot be explained by increasing DF ROM or PF weakness alone. The change in peak ankle DF TST prior to age 13 may be the result of an interaction between PF weakness, PF tightness, increasing body weight and changing foot structure which allows more dorsiflexion in stance with age and enables the existing ankle dorsiflexors to function more effectively in swing. Consistent peak dorsiflexion in TST after 13 years of age suggests that disease progression may not confound the understanding of treatment results in older children. Similarly, consistent plantarflexion in MSW after age 13 suggests that “drop foot” is less likely to develop. Patients with CMT2 showed no change with age, a younger onset (prior to age 5) and more variation in ankle deviations. It is likely that the disease progression occurred earlier than the initial gait assessments. The excessive DF in TST and plantarflexion in MSW suggest the need for more bracing in CMT2 vs. CMT1. Heterogeneity in ankle function in CMT2 highlights the importance of understanding disease progression at an individual patient level.

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The Effectiveness of Serial Casting in the Treatment of Recurrent Equinovarus in Children with Arthrogryposis

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INTRODUCTION

Arthrogryposis multiplex congenita (AMC) is characterized by joint contractures in two or more body areas, often with resulting foot deformities [1]. The most common foot deformity in AMC is clubfoot (equinovarus) [1]. This type of clubfoot is typically stiffer than an idiopathic clubfoot and has traditionally been treated surgically, despite high rates of recurrence and complications [2]. In contast, idiopathic clubfoot is successfully treated with serial casting, resulting in improved foot mobility while reducing the number of invasive surgical procedures required [3]. Recently, serial casting has been encouraged for the stiffer clubfoot seen in AMC, despite little research to support its use. The existing literature suggests the Ponseti method, which involves casting, orthotics and physical therapy, could be useful in treating children with arthrogrypotic clubfeet, but no information is available regarding recurrence rates, need for further surgical intervention, or functional implications of the correction.

CLINICAL SIGNIFICANCE

The clubfoot in children with AMC has traditionally been treated surgically [2], but recently conservative management with serial casting has been encouraged. Improved outcomes with conservative treatment could reduce the need for invasive surgical procedures.

METHODS

Children with AMC treated with serial casting to address persistent or recurrent clubfoot were evaluated retrospectively comparing outcomes before, short term (within 6 months, ST) and longer term (between 6 and 14 months, LT) after casting. Outcomes included ankle and foot passive range of motion (PROM), dynamic foot pressure (Tekscan, Boston, MA USA), parent reported Pediatric Outcomes Data Collection Instrument (PODCI), brace tolerance (good=no pain with regular use, fair=pain or redness with use, poor=pain or breakdown limits brace use), and the need for post-casting surgery. ANOVAs or paired t-tests were used as appropriate to determine the effects of casting. Changes in brace tolerance before and after serial casting were analyzed using the Global Test for Symmetry[4].

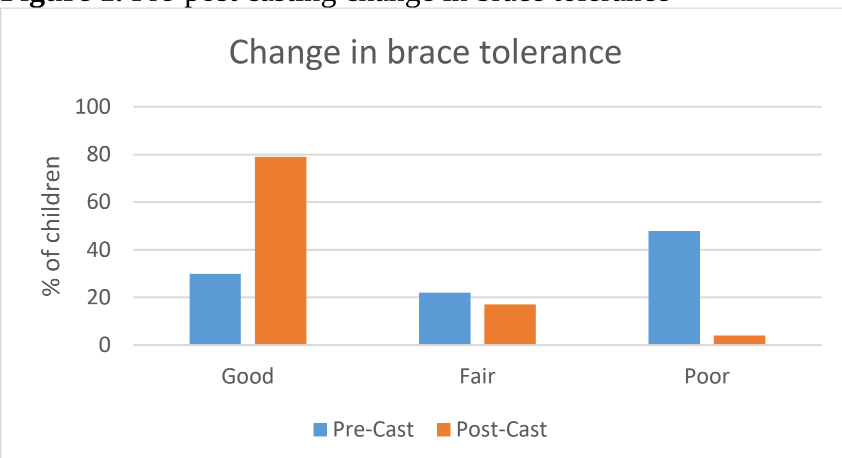
RESULTS

Fifty children (6.3±3.5 years old) were casted 2.5±1.9 times, with a total of 214 serial casting episodes analyzed. Children were casted weekly for a period of 26±11 days followed by ST assessment 0.3±1 months post-casting, and LT assessment 8.5±2.7 months post-casting. Sixty-eight percent of children used AFOs, 28% KAFOs, and 4% in shoe or no orthoses. PROM showed improvement in ankle dorsiflexion and forefoot abduction in ST, returning to baseline measurements in LT (Table 1). Brace tolerance improved after casting (pre: good 30%, fair 22%, poor 48%, post: good 79%, fair 17%, poor 4%; $p<0.05$; Figure 1). With follow-up to 10.3±5.5 years of age, only 20% of feet required surgery. For those that required surgery, it was completed 4.7±3.2 years after casting was initiated. There were no significant changes in dynamic foot pressure, or PODCI results after serial casting except for an increase in the pain subtest ($p<0.05$).

Table 1: Comparison of PROM at pre, ST, and LT post serial casting

	Pre		ST		LT		Pre vs ST	Pre vs LT	ST vs LT
	Mean	SD	Mean	SD	Mean	SD	P	P	P
Ankle Dorsiflexion PROM n=120	-11.0	12.1	-2.9	8.8	-8.4	11.1	1.92E-11	0.105	1.47E-.07
Forefoot Abduction PROM n=65	7.6	17.2	19.3	11.3	6.8	14.6	1.78E-06	0.9	2.54E-10

Figure 1. Pre-post casting change in brace tolerance



DISCUSSION

Traditionally foot deformities in children with arthrogryposis have been treated with surgical intervention, due to the rigidity and severity of the abnormal alignment. Serial casting in children with arthrogryposis is effective in improving PROM in the short term, but baseline measures recur in the longer term. Serial casting improves brace tolerance and delays the need for invasive surgical procedures in children with arthrogryposis.

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THE EFFECT OF ANKLE-FOOT-ORTHOSIS ON LOADING OF THE UNINVOLVED EXTREMITY IN CHILDREN WITH HEMIPLEGIC CEREBRAL PALSY

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INTRODUCTION

Children with spastic hemiplegic Cerebral Palsy (SHCP) develop gait abnormalities secondary to muscle imbalances and spasticity. However, the gait abnormalities result in abnormal loading patterns at different joints. At the hip, with the use of computer simulations, abnormal cyclic loading patterns during walking have been shown to affect the morphology of the joint, especially during the years of skeletal development [1]. Conditions of normal loading at the hip result in growth areas rotating posteriorly, decreasing the anteversion angle and approaching normal adult angles. Conversely, the loading patterns of children with CP result in the femoral growth area rotating anteriorly, increasing the anteversion angle [2].

To account for the abnormal loading patterns children with SHCP, and CP in general, are often referred for orthotic treatment. Treatment goals for orthotic management of SHCP are: (a) to correct and/or prevent deformity; (b) to provide stability; (c) facilitate activities of daily living; (d) improve gait efficiency; and (e) approximate normal gait [3]. Solid ankle and articulated ankle foot outhouses (AFO) are used for children who are able to achieve some level of walking ability in order to achieve these goals [4]. Previous studies have focused on the kinetic and kinematic effects of un-braced and braced conditions on the involved limb in children with SHCP, however, few have looked at the effect on the uninvolved limb [5].

The purpose, therefore, of this study was to investigate the loading patterns of the uninvolved hip in children with SHCP by comparing braced and un-braced conditions.

CLINICAL SIGNIFICANCE

By focusing on the mechanics of the uninvolved limb during ambulation, we may gain an understanding of how the braced limb affects the uninvolved limb and in turn help further understand how orthotic treatment affects the body as a whole. The use of an orthosis on the involved side may affect the forces and moments through the uninvolved limb. As the gait pattern approaches normalcy with the use of an AFO, abnormal forces through the developing hip joint may be normalized as well.

METHODS

This retrospective study included 16 children with SHCP, 6-18 years of age (average 9 years old), Gross Motor Functional Scale (GMFCS) II. Subjects were excluded if they had undergone any surgery within 12 months from the time of data collection or if they were using any other ambulatory device. No participant had history of other neuromusculoskeletal diagnosis. At our hospital's Motion Analysis laboratory, all subjects underwent three-dimensional gait analysis. Subjects walked at self-selected pace, barefoot and with their prescribed AFO on the involved LE.

Maximum anterior/posterior forces and transverse plane moments at the involved and uninvolved hips, along with the respective timings, were calculated. All data were normalized. Timing was expressed as percent of the gait cycle. Forces and moments were expressed relative to the subject's body weight.

RESULTS

Forces were not statistically different between the involved and uninvolved hips when walking with the AFO. Posterior forces at the involved hip increased when walking with the AFO compared to barefoot (p=0.01). The same was true for the moments.

Table 1: Maximum Barefoot (BF) vs Brace (AFO) anterior (Ant) and posterior (Post) forces for the involved (I) and uninvolved (U) hip. Statistically significant results are italicized.

	Mean	**Diff.**	**p**			**Mean**	**Diff.**	**p**
AntIBF **AntIAFO**	2.443 2.552	-0.1087	0.411		**AntIBF** **AntUBF**	2.443 2.316	0.126	0.812
PostIBF **PostIAFO**	*-1.789* *-2.527*	*0.737*	*0.011*		**PostIBF** **PostUBF**	-1.789 -2.293	0.503	0.174
AntUBF **Ant UAFO**	2.316 2.454	-0.137	0.325		**AntIAFO** **AntUAFO**	2.552 2.454	0.098	0.841
PostUBF **PostUAFO**	-2.293 -2.447	0.154	0.587		**PostIAFO** **PostUAFO**	-2.527 -2.447	-0.079	0.860

DISCUSSION

To the best of our knowledge, this is the first study investigating the loading effects on the uninvolved hip as a function of walking with and without an AFO. The results demonstrate symmetrical loading of the involved and uninvolved hips when walking with the AFO. Consequently, our results suggest that the use of AFOs is an efficacious and appropriate treatment when considering the loading patterns at the hips of the involved and uninvolved lower extremities.

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DISCLOSURE STATEMENT

The authors have nothing to disclose.

IMMEDIATE AND 6-WEEK EFFECTS OF USING CARBON FIBER FOOTPLATES ON GAIT IN CHILDREN WITH IDIOPATHIC TOE-WALKING

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INTRODUCTION

Idiopathic toe walking (ITW) is a phenomenon of unknown etiology that can affect children as young as 18 months and can persist into adolescence [1,2]. Children with ITW have significant gait deviations that include, but are not limited to, decreased dorsiflexion during initial contact, decreased hip extension during toe off, decreased stride length, and overall limited time in stance phase when compared to typically walking children of the similar age [3]. Previous studies have also explored differences in joint kinematics and kinetics between children with ITW and children with cerebral palsy [4].

Carbon fiber footplate orthoses (CFOs) can be prescribed as an intervention for children with ITW to biomechanically restrict the child from rising up onto their forefoot throughout stance phase. There is limited literature on estimating the immediate and long-term effects of using CFOs on gait in children with ITW [5]. The purpose of the current study was to: 1) Explore differences in spatio-temporal and center of pressure-based gait parameters in children with ITW and children with typical gait pattern and 2) Examine the immediate and 6-week effects of using CFOs on gait on children with ITW.

CLINICAL SIGNIFICANCE

ITW is a difficult condition to treat. CFOs are an affordable and convenient intervention with potentially positive impacts on gait in children with ITW. Benefits to clinicians include providing conservative and inexpensive treatments before pursuing more aggressive options.

METHODS

34 children between ages 3 to 9-years old participated. Among them, 20 were children who idiopathically walked on their toes (ITW group; ITWG) and 14 were children who walked with a typical gait pattern (Control group; CG). None of the children had any neurological diagnosis, surgical history or used current orthoses. Each participant walked at a normal pace on a 16’ long x 4’ wide instrumented pressure walkway (Zeno Walkway, Protokinetics LLC, Havertown, PA) for 5 laps wearing their regular shoes (PRE). Participants in the ITWG then walked for 5 more laps with CFOs inserted beneath bilateral insoles (IMMEDIATEPOST). The ITW group was then prescribed to wear the CFOs for six weeks during typical activities, after which their walking was again assessed similar procedures (6WKSPOST). Spatial gait parameters like stride length and width, and temporal gait parameters like % of stance, swing, single support and double support phases, gait speed and cadence were extracted and used for analyses. In addition, center of pressure characteristics like the distance travelled by the center of pressure during stance phase, single support phase and double support phase were also extracted. An average of all the steps from the 5 passes was used, and values were averaged between right and left legs. Independent samples t-tests were done to compare the two groups. A one-way repeated measures ANOVA was performed to compare the immediate and prolonged effects of CFO on gait within the ITWG.

RESULTS

When walking without CFOs, children in the ITWG had significantly lesser stance phase %, lesser double support phase %, and lesser center of pressure distance during single support phase (Table 1). These differences did not exist when comparing gait kinematics of ITW walking with CFOs with those of CG. Across time, significant differences were primarily seen between IMMEDIATEPOST and PRE and IMMEDIATEPOST and 6WKSPOST with greater stance phase %, and lesser single support phase % during IMMEDIATEPOST (Table 1). There were no other significant differences.

Table 1: Mean (SD) of gait variables that showed significant differences

Gait parameter	PRE-CG	PRE-ITWG	IMMEDIATEPOST-ITWG	6WKSPOST-ITWG
Stance phase %	60.16 (1.57)	59.21 (1.12)	59.99 (1.35)	59.01 (1.09)
Single support phase %	39.83 (1.33)	40.69 (1.14)	39.97 (1.49)	41.04 (1.08)
Double support phase %	20.47 (2.50)	18.52 (2.23)	20.00 (2.85)	17.96 (2.18)
CoP distance during single support phase (cm)	11.70 (2.00)	9.24 (2.72)	11.12 (3.15)	10.20 (1.97)

DISCUSSION

Children who idiopathically walk on their toes seem to achieve similar spatial gait characteristics and functional outcomes (like gait speed and cadence) but different temporal gait characteristics compared to children with typical gait pattern. Center of pressure movement which is reflective of weight-bearing was primarily altered during the single support phase. Immediate effects of the CFOs seem to lessen these differences. Further, usage of CFOs seems to alter the temporal gait pattern immediately but not after 6 weeks of usage. Similar with immediate improvements in temporal measurements were observed when using CFOs compared to walking barefoot [5]. Impact of longer CFO usage needs to be evaluated.

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DISCLOSURE STATEMENT

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Non-Linear Ankle Foot Orthosis Stiffness in Musculoskeletal Models

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INTRODUCTION

Ankle Foot Orthoses (AFOs) are assistive devices prescribed to correct pathologic gait patterns found in several patient populations such as stroke survivors and children with cerebral palsy. AFOs alter gait kinetics by adding additional torques onto the ankle joint. While the effect of AFO stiffness on pathologic gait is an actively pursued research question, current studies using musculoskeletal models to characterize pathologic gait and model the effects of AFO interventions use a simplified linear AFO model to represent AFO torque [1]. To improve the accuracy of AFO modeling, we propose a new method for modeling non-linear AFO stiffness, describing AFO torque response as a function of ankle angle.

CLINICAL SIGNIFICANCE

Most commercially available AFOs produce a non-linear torque response to deformation along the primary axis of the ankle. Accurate implementations of non-linear AFO torque response in musculoskeletal models increase their viability as a tool for AFO prescription.

METHODS

The stiffness profiles of a double action ankle joint were determined given the joint's geometry and the experimentally measured spring constant of the device's internal springs for multiple set screw depth settings (Figure 1). Three healthy subjects were instructed to perform passes across a 14 meter force plate embedded walkway at a self-selected normal speed. We attached reflective markers following a modified Helen-Hays set to record the passes with a motion capture system. The subjects performed 10 passes for 6 conditions: walking while wearing the AFO on the right foot with the set screws 0mm, 4mm, 8mm, 12mm, and 16mm into the device and normal walking (without the AFO). Additional passes were added until at least 5 good passes over force plates were achieved.

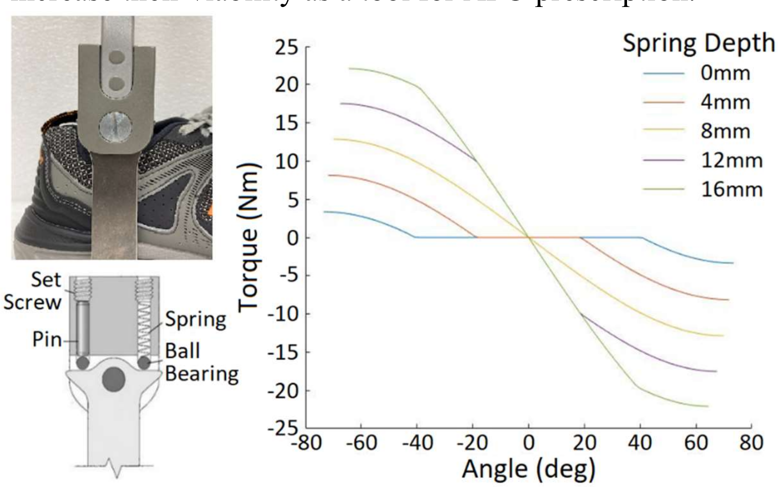

Figure 1: The double action ankle joint and a schematic of its internal mechanism (Left). The theoretical torque profile of the double action joint with various set screw depths (Right).

Inverse kinematics was performed on a scaled model developed by Rajagopal et al. in OpenSim [2]. The torque generated by the AFO was determined based on the theoretical torque angle curve and the ankle angle data produced by inverse kinematics. These torques were then applied to the talus and tibia segments as an external force. Inverse dynamics was performed on the models using the gait kinematics, ground reaction forces, and AFO torques.

DEMONSTRATION

Typical plots of ankle angle and corresponding theoretical AFO torque production are shown in Figure 2. While ankle kinematic extrema diminish in magnitude with deeper set screws and more AFO stiffness, the maximal AFO torque production remained similar.

As shown in Table 1, the kinematics and kinetics of gait without the AFO (subjects walking in their own shoes) differed from those of gait with the AFO without springs. Peak ankle dorsiflexion changed significantly for subjects 1 and 2 when the springs were added, while the peak internal ankle moment changed significantly for subjects 2 and 3. The internal ankle moment and the AFO's applied torque are combined to find the net ankle moment. No significant differences were found in net ankle moment for each of the AFO conditions.

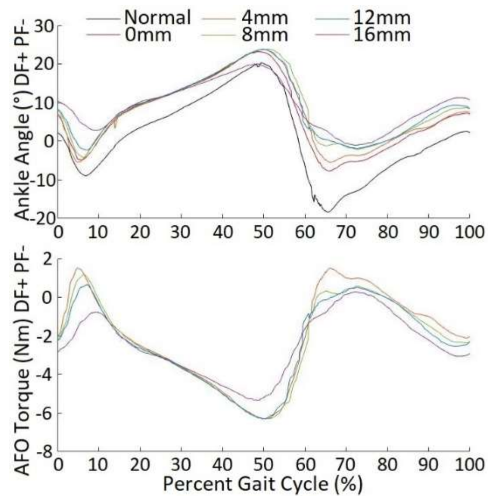

Figure 2: Ankle angles of subject 3 with various AFO conditions (Above). The corresponding AFO torques (Below).

Table 1: Table shows peak ankle: dorsiflexion angle, internal moment, and net moment. Values outside one standard deviation of the 0mm condition value are highlighted in red.

Condition	Peak Dorsiflexion (°)			Peak Internal Ankle Moment (Nm/Kg)			Peak Net Ankle Moment (Nm/Kg)		
Subject	1	2	3	1	2	3	1	2	3
Normal	19.00 ±0.95	16.55 ± 0.27	20.34 ±0.62	-1.03 ±0.02	-1.47 ±0.03	-1.42 ±0.05	-1.03 ±0.02	-1.47 ±0.03	-1.42 ±0.05
0mm	21.32 ±0.32	21.81 ±0.18	23.26 ±0.48	-1.01 ±0.06	-1.57 ±0.01	-1.32 ±0.03	-1.01 ±0.06	-1.57 ±0.01	-1.32 ±0.03
4mm	19.81 ±0.64	20.25 ±0.88	23.77 ±0.83	-1.04 ±0.03	-1.55 ±0.05	-1.15 ±0.11	-1.12 ±0.05	-1.62 ±0.05	-1.23 ±0.11
8mm	21.43 ±0.57	19.25 ±0.19	23.86 ±1.00	-1.00 ±0.03	-1.49 ±0.03	-1.14 ±0.07	-1.08 ±0.04	-1.56 ±0.03	-1.27 ±0.07
12mm	19.83 ±0.74	20.34 ±0.51	23.89 ±0.54	-0.98 ±0.04	-1.50 ±0.02	-1.30 ±0.04	-1.05 ±0.05	-1.57 ±0.03	-1.38 ±0.04
16mm	13.60 ±0.75	19.31 ±0.30	20.01 ±0.69	-0.88 ±0.04	-1.50 ±0.02	-1.19 ±0.03	-0.93 ±0.04	-1.57 ±0.02	-1.27 ±0.03

SUMMARY

Results of this study indicate that healthy subjects adapt to gait with the various AFO ankle stiffness settings by altering their ankle kinematics in order to preserve their net ankle kinetics, a finding consistent with previous literature [1]. This validates the presented method for modeling non-linear AFO stiffness as it was successfully implemented in musculoskeletal models to reproduce a well-studied phenomenon regarding variable AFO stiffness.

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DISCLOSURE STATEMENT

Authors have no conflicts of interest to disclose.

COMPARISON OF THREE AFO DESIGNS ON WALKING FUNCTION, AVERAGE DAILY STEPS, AND AFO PREFERENCE IN INDIVIDUALS POST STROKE

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INTRODUCTION

There are approximately 795,000 strokes per year in the United States, with 550,000 surviving the acute phase [1]. Although approximately 60 to 95% of stroke survivors will recover the ability to independently walk short distances without assistance of another person, only 50% achieve even limited community ambulation [2]. In addition to corrective physical therapy, ankle-foot orthoses (AFOs) are frequently prescribed for patients with hemiplegia to enhance walking function [3]. However, choosing an effective orthosis that is affordable and agreeable to the patient is largely a trial-and error process because our understanding of the clinical indicators for when an off-the-shelf (OTS) semi-rigid AFO versus a custom-made, articulating AFO will optimize walking are limited. Because the cost of custom-made poly articulating AFOs are typically 3-5 times greater than OTS semi-rigid AFOs, selection of an AFO is complicated by health-care related financial pressures that drive clinicians to prescribe the less expensive OTS AFOs. However, an AFO that does not address all of an individual's critical impairments and preferences is not only economically wasteful, it can actually impair walking function further [3] and could hinder full participation in the home and community. The aim of this ongoing study is to identify the clinical criteria for determining which of three commonly used AFO types would be most beneficial for individuals with hemiplegia after stroke: two semi-rigid OTS designs made of either polypropylene (POLY) or a carbon composite (CC) and a custom-made polypropylene articulating AFO with a dorsiflexion stop and dorsiflexion spring assistance (DSDA).

CLINICAL SIGNIFICANCE

Identifying clinical indicators for an evidence-based, decision-making AFO prescription guideline for persons with hemiplegia from stroke can benefit those who require a lower extremity AFO for daily ambulation in the community.

METHODS

Ten participants greater than three months post stroke who had been prescribed an AFO and walked between 0.2 – 1.0 m/s were randomly tested in each of the three different AFOs: POLY, CC, DSDA (Fig. 1). Prior to testing, participants wore each AFO for two weeks. A StepWatch™ activity monitor was attached to the AFO to track daily stepping activity. During each lab test, participants walked across a 6-meter walkway at their self-selected free and fast walking speeds and completed a Six-Minute Walk Test (6MWT). For the 6-meter tests, gait velocity and stride characteristics were determined using compression closing foot switches taped to the bottom of each shoe. After the final testing session participants were asked to select their preferred AFO.

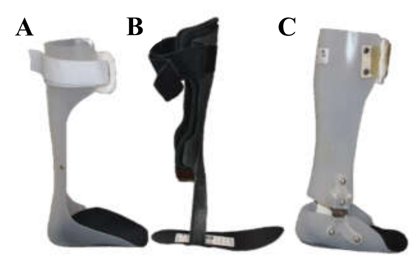

Figure 1: Study AFO Designs: A) OTS Semi-rigid Polypropylene (POLY), B) OTS Semi-rigid carbon composite (CC), C) Custom –made articulated polypropylene (DSDA)

RESULTS

The 10 participants were male with an average age of 51.5 years (35.8 – 62.2)and a time post stroke of 11.9 months (4.4 – 30.2). Mean velocity and stride length were greatest using the DSDA AFO in both free and fast walking in 7 participants (Table 1). The participants also had the greatest distance walked during the 6MWT (Table 1). Daily number of steps walked in each brace was variable and not correlated with either the preferred brace or the brace in which they were fastest. Five participants walked more steps per day in the CC AFO, four in the POLY AFO, and one in the DSDA AFO. Preferred brace was not consistent among participants: 3 selected POLY, 4 CC, and 3 DSDA. The most common reasons reported for selecting the preferred brace in participants who chose the CC or POLY AFOs were lightness and ease of donning/doffing. Those who chose DSDA AFO reported more normal movement at the ankle and that it felt "stronger" during walking.

Table 1: Walking metrics and preferred AFO for 10 participants. Mean velocity and stride length data are shown for free walking only. Fast walking results were qualitatively similar.

	Free Walking						6MWT			StepWatch			
	POLY		CC		DSDA		POLY	CC	DSDA	POLY	CC	DSDA	
Subject	Vel (m/sec)	Stride Length (m)	Vel (m/sec)	Stride Length (m)	Vel (m/sec)	Stride Length (m)	Distance (m)	Distance (m)	Distance (m)	Avg steps/day	Avg steps/day	Avg steps/day	Chosen AFO
1	0.72	1.154	0.8	1.185	0.83	1.215	307	304	325	2166	877	1502	POLY
2	1.10	1.094	0.76	0.953	0.86	0.967	435	385	395	8111	4047	6080	CC
3	0.77	1.102	0.97	1.233	0.66	1.035	330	349	331	1724	2719	634	CC
4	0.25	0.461	0.23	0.435	0.27	0.505	75	69.5	80	2233	1703	1858	POLY
5	0.66	1.017	0.74	1.074	0.84	1.168	372	365	391	4601	4504	4001	DSDA
6	0.38	0.594	0.42	0.607	0.47	0.761	209	219	271	no data	2947	2612	DSDA
7	0.62	0.941	0.63	0.933	0.74	1.03	240	245	255	2608	4269	6366	DSDA
8	0.75	1.014	0.66	0.985	0.75	0.998	270	255	248	0	840	0	CC
9	0.89	1.151	0.9	1.177	1	1.249	299	292	318	501	539	273	POLY
10	0.80	1.018	0.78	0.978	0.93	1.103	300	299	287	1322	3529	2985	CC

DISCUSSION

All ten participants demonstrated improved walking function when wearing an AFO compared to walking in shoes only. Despite greater improvement in walking function (step length, velocity, and distance) using the custom articulating AFO (DSDA) in 7/10 participants, there was no difference in AFO usage in the home and community as measured by daily step count. Nor was there a correlation between walking speed and brace preference. Ease of use (orthosis weight and ease of donning and doffing) may be as important a criterion to consider as gait function improvements when selecting the appropriate AFO.

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DISCLOSURE STATEMENT

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DIFFERENCES IN PELVIS KINEMATICS USING ILIAC CREST MARKERS COMPARED WITH ANTERIOR SUPERIOR ILIAC SPINE

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INTRODUCTION: In many motion analysis protocols, movement of the pelvis is tracked using markers on the anterior superior iliac spines (ASIS). However, visibility of the ASIS markers is sometimes problematic for subjects with excessive adipose tissue or during motions involving large hip flexion like the drop vertical jump. In these cases, markers with more lateral placement, such as on the iliac crest (IC), are often used for gap filling or tracking. It is unknown whether these markers provide equivalent kinematics to ASIS markers. Therefore, the purpose of this study was to compare pelvis and hip kinematics determined using IC versus ASIS markers.

CLINICAL SIGNIFICANCE: On average, only small differences were observed when tracking with IC vs. ASIS markers, suggesting that they can be used interchangeably. While larger differences are sometimes observed for individual patients, a gold standard such as imaging would be needed to determine which markers provide more accurate tracking.

METHODS: Pelvis and hip kinematics were calculated for 10 patients post-anterior cruciate ligament reconstruction (7 male; age 15-19 yr; BMI 17.5-31.1 kg/m^2) performing walking at a self-selected pace, drop vertical jump from a 41 cm step, and 45° cutting. Three models for tracking the pelvis were compared. All models used the same segment definitions for the pelvis and thigh based on the conventional gait model. The thigh segment was defined using the Harrington hip joint center and medial/lateral markers on the knee joint axis. The pelvis was defined using markers on the left and right ASIS and a marker on the sacrum (SACR). The thigh was tracked using a rigid cluster on the anterior distal thigh. The pelvis was tracked using the sacral (SACR) marker and either the ASIS markers (LASI, RASI), IC markers (LIC, RIC), or both (LASI, RASI, LIC, RIC). All modeling was performed in Visual3D v2021 (C-Motion Inc., Germantown, MD). Kinematics of the pelvis (relative to the lab) and hip (thigh relative to pelvis) were extracted for one right limb cycle per subject and task and averaged across all subjects. The difference in average curves and their range of motion (ROM) were compared between the ASIS and IC models. The root mean squared difference (RMSD) was also calculated for each subject, and the relationship between RMSD and body mass index (BMI) was examined using Spearman rank correlation.

RESULTS: On average, pelvis and hip angles differed slightly between IC and ASIS (Fig. 1), but the mean magnitude of difference was <1° for all angles during the walk and <3° for the drop jump and cut (Table 1). IC produced smaller ROM than ASIS in the coronal plane for both the pelvis and hip, but greater hip rotation ROM during all motions. Differences were sometimes greater for individual subjects, with RMSD up to 6-10° in some cases. RMSD was higher for patients with greater BMI for pelvis rotation and hip ab/adduction during the drop jump and pelvis and hip rotation during walking ($r>0.64$, $p<0.05$).

Table 1: Summary of differences between IC and ASIS (IC – ASIS, deg)

	Average Curves						Individual Subjects					
	Mean Difference			Difference in ROM			RMSD Mean			RMSD Max		
	Walk	DJ	Cut	Walk	DJ	Cut	Walk	DJ	Cut	Walk	DJ	Cut
Pelvis X	0.7	2.8	1.4	0.4	0.6	0.5	1.0	4.9	2.9	2.4	9.6	7.0
Pelvis Y	0.1	0.2	-1.3	-4.7	-2.8	-3.1	1.6	1.7	3.4	2.1	2.5	6.8
Pelvis Z	-0.3	-0.7	-1.4	-2.5	-1.4	0.8	1.4	1.6	3.4	2.1	2.9	5.5
Hip X	0.7	2.9	1.6	-0.7	-0.2	0.1	1.0	5.0	3.1	2.3	9.7	8.6
Hip Y	0.4	0.7	0.5	-3.5	-2.5	-2.6	1.5	1.4	2.2	2.2	2.9	5.3
Hip Z	0.6	0.2	2.1	4.5	3.2	5.9	1.6	1.9	4.4	2.6	3.	7.4

X=sagittal, Y=coronal, Z=transverse

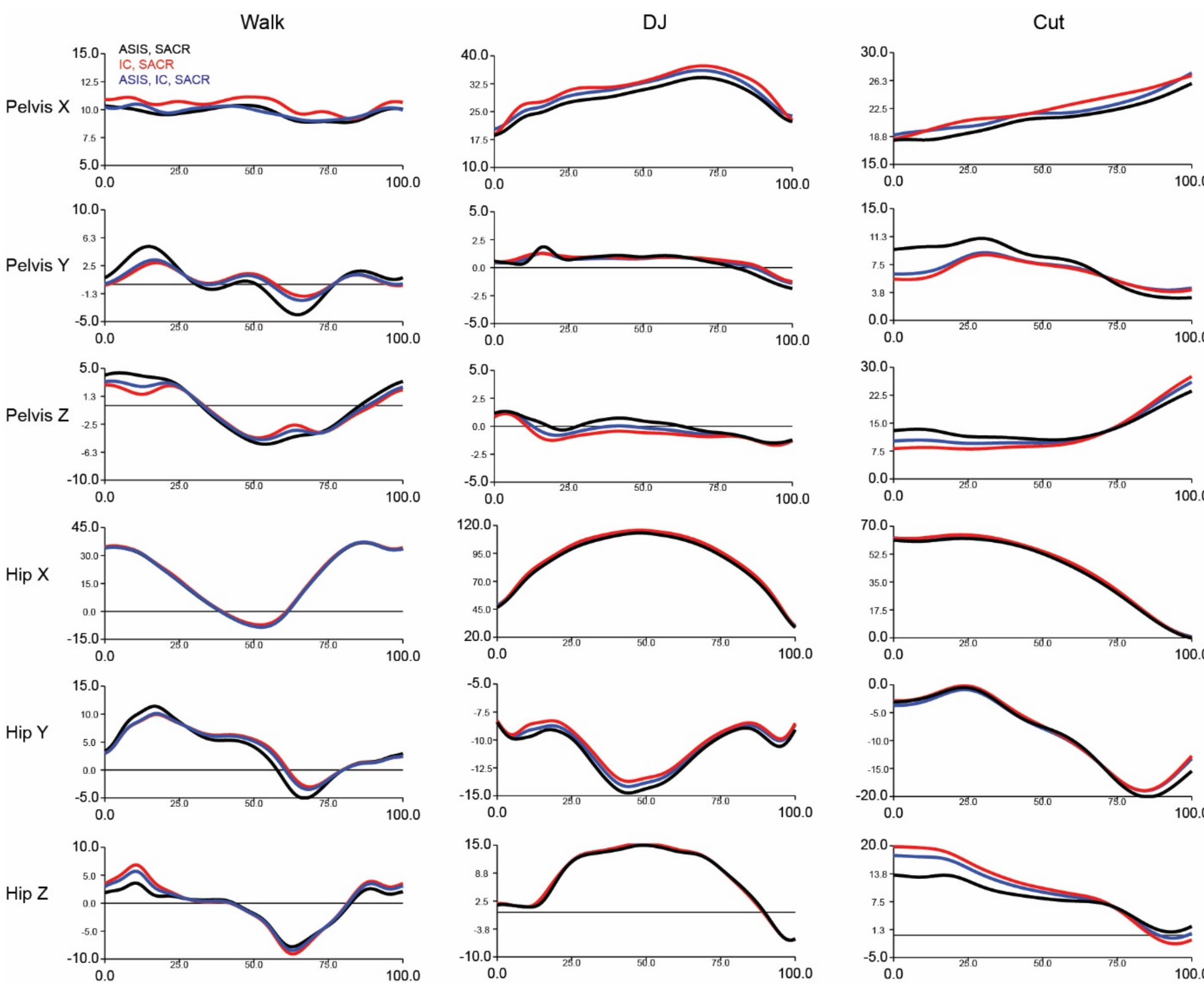

Figure 1: Average curves obtained using different tracking markers

DISCUSSION: There was little systematic difference in kinematics using IC vs. ASIS for tracking, but individual subjects can show greater differences. Without a gold standard such as imaging, we cannot determine which model provides more accurate kinematics. Practitioners should be able to use either marker set while being aware of these differences.

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Simple Check of Knee Flexion Extension Axis

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INTRODUCTION

This paper introduces a simple functional measure as a check of the knee flexion extension axis (KFEA). Most clinical motion labs present knee motion as the relative 3D rotation of the lower leg with respect to the thigh given local coordinates embedded in each segment[1]. The relative motion is presented as a time series of clinical angles generated from mathematical calculations (ex. Euler, Helical). Regardless of the mathematical calculation utilized, the results depend on the perspective or the determination of the reference axes for the knee, specifically the KFEA. KFEA is determined by simple anatomical models based on visual estimation (i.e. therapist places markers on femoral epicondyles[2, 3]), mechanical devices (Knee Alignment Device (KAD[4]) or functional calculations[5, 6]. Given that there are limitations in current models and the nature of how these modes describe movement, and there are additional measurement errors (e.g. marker placement inaccuracies) that compound these errors, a simple functional check would be prudent.

CLINICAL SIGNIFICANCE

By simply plotting the motion of the ankle joint center in the reference frame of the thigh which includes the determined KFEA one can easily determine the accuracy and the utility of the kinematic knee model as applied to the individual subject that is being analyzed.

METHODS

Calculate the imbedded local thigh coordinate system using whatever anatomical, mechanical, or functional method you typically utilize. Calculate the motion of the ankle joint center from the perspective of the thigh segment during the walking trials of your data collection. I specify the use of walking trials as opposed to a passive test as, moments and forces of weight bearing walking will make the motion of the knee different from a passive motion. Additionally there may be marker motion due to skin, adipose, and muscle motion (artifact) during walking that differs from that in passive motion. Plot the motion of the ankle as a 2D projection of flexion-extension and varus-valgus. In an ideal situation of a pure well aligned hinge joint the ankle plot will be along a single vertical line with no varus-valgus bias. In practice, this will never occur as the knee does not function as a pure hinge joint. That is, the KFEA is not fixed and moves and rotates with respect to the fibula[7, 8]. In addition there will be an error in the determination of the ankle joint's relative positon to the thigh due to errors in the optical measurement of the markers and marker motion artifact.

DEMONSTRATION

Figure 1(A) are the knee kinematic plots of a normal individual with (B) graphs of the ankle joint position in the thigh's coordinate system. The numbers on the right side of the ankle joint center plot are angular deviation of a best fit line through the ankle position data from the vertical (perpendicular to KFEA).

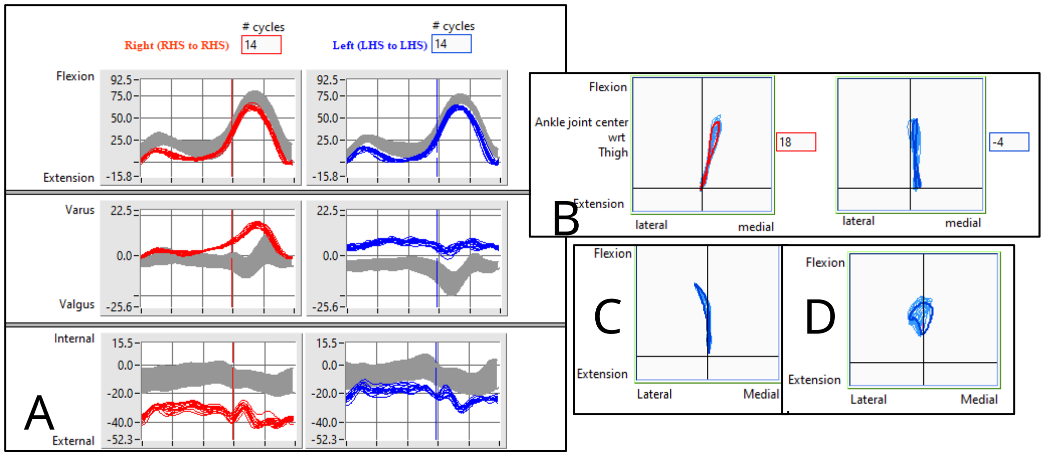

Figure 1 (A) knee kinematics (B-D) ankle in thigh coordinate system

As one can see, the left KFEA is fairly well aligned, for the path of the ankle joint only deviates from vertical by 4^{o}. The right side shows a deviation of 18^{o} from pure flexion extension. This is also reflected in the cross talk seen in the varus-valgus kinematic graphs. Figure 1(C and D) show examples of the knee joints of other individuals not acting as the idealized 1D hinge. The first (1C) depicts an individual with pure flexion/extension in stance and a lateral deviation in swing and the second (1D) depicts an individual with "fixed-knee" gait that is in crouch, but there is an almost equal medical-lateral component. In total the motion is more of a low amplitude circumduction. Thus, given forces present in walking, geometry and orientation of the structures the knee and laxity in the joint may actually function along a direction that is not congruent with the anatomy.

SUMMARY

Regardless of the method of KFEA determination or kinematic calculation a simple check of the ankle joint motion in the thigh coordinate system provides valuable insight to the accuracy, utility of the calculations and model and provides a clear visualization of the clinical function of the knee. Since this calculation is so simple to do there really is no reason not to do it.

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DISCLOSURE STATEMENT

John Henley has no conflicts of interest to disclose.

IT'S ALL IN THE MOMENT

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BACKGROUND AND PATIENT HISTORY

Patients with complaints of anterior knee pain, patellar instability and/or gait dysfunction who have failed conservative treatment are often referred by orthopedists for instrumented gait analysis to quantify the transverse plane mechanics which might underlie their patient's pain/instability/dysfunction. External tibial torsion is a common feature among such patients. Motion capture findings are variable, but often present is an atypical knee varus moment with running that is absent in walking. A case example is provided: AH is a typically developing (TD) 10+4-year-old male who came to a clinical motion laboratory with concerns that his right foot "points out" when walking and running, which was felt to negatively impact his walking and involvement in the sports of baseball, basketball, soccer, and taekwondo. He had no pain complaints.

CLINICAL DATA

Gait by observation documented independence with ambulation, with minimal visual deviations in any plane of motion, except for asymmetric foot progression with the right more external than left, both in walking and running. Exam findings demonstrated the presence of right external tibial torsion, with a bimalleolar axis (BM) of 28 degrees. There was no clinical evidence for excessive femoral anteversion, genu varum/valgum, or leg length inequality.

MOTION DATA

Transverse plane kinematic data demonstrated knee flexion-to-bimalleolar axis knee rotations greater than 2 standard deviations external on the right, both when walking (blue, Fig.1) and running (blue, Fig. 2). *Coronal* plane knee kinematics were normal in walking (Fig. 3) and running (Fig. 4).With walking, *coronal* plane knee kinetics were within typical limits (Fig. 5). However, with running, AH developed an inappropriate knee varus moment during single support (Fig. 6).

WALKING: pre/post

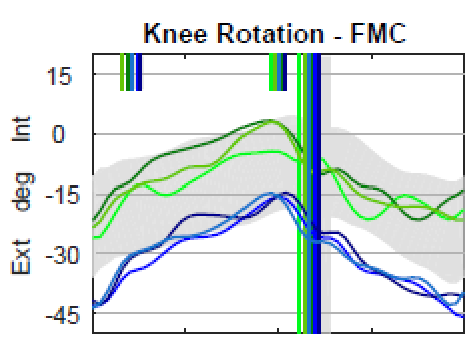


Fig. 1

RUNNING: pre/post

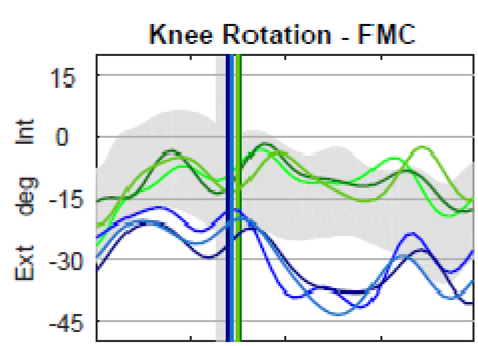
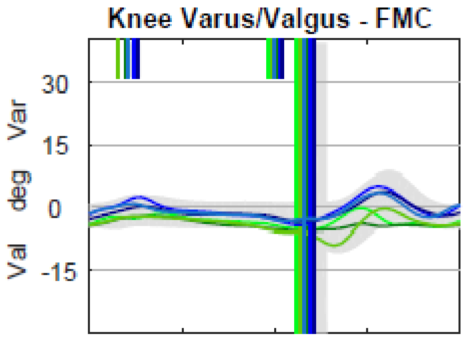


Fig. 2



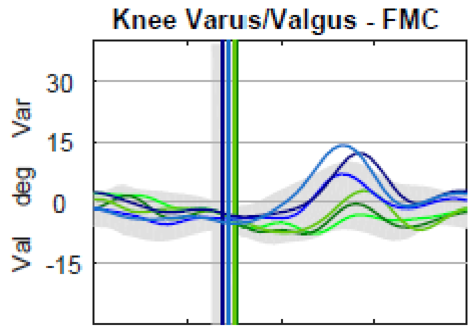
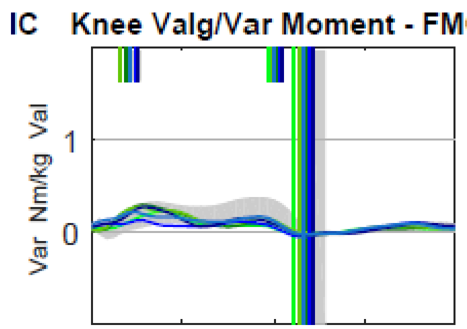


Fig. 3 **Fig. 4**



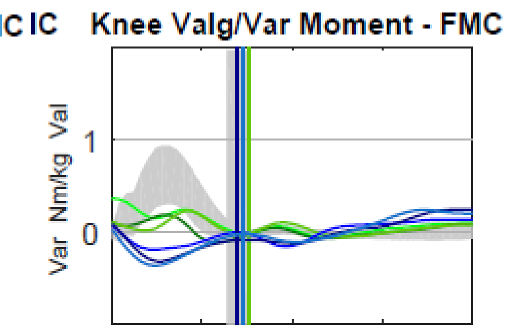


Fig. 5 **Fig. 6**

TREATMENT DECISIONS AND INDICATIONS

Given the family's subjective concerns, in conjunction with motion data confirming the presence of isolated external tibial torsion, it was felt that AH's functional symptoms were most likely related to external tibial torsion. AH underwent a right distal internal tibial de-rotation osteotomy of approximately 23 degrees.

OUTCOME:

Subjectively, AH and his family reported being "very pleased" with results of surgery, with improvement noted in "coordination and running, with better ability to keep up with peers". Transverse plane right knee rotations normalized post operatively with walking and running (green, Fig. 1,2), with an improved/smaller magnitude knee varus moment in running (Fig. 6) that continues to be absent with walking (Fig. 5).

SUMMARY

This case study illustrates the use of instrumented gait analysis to identify isolated external tibial torsion which was felt to negatively impact an older child's gait and function and contributed to generating an atypical varus knee moment in running. This has been noted in approximately 73% of TD patients queried who have confirmed external tibial torsion in the absence of coronal plane knee malalignment. Questions that remain are why this atypical knee moment occurs in running but not walking, and how this atypical running moment may contribute to knee pain, patellofemoral dysfunction, and/or gait dysfunction in a high functioning, TD population.

DISCLOSURE STATEMENT

S. Sohrweide and A. Georgiadis have no conflicts of interest to disclose.

QUANTITATIVE ASSESSMENT OF DYNAMIC LIMB LENGTH AND TOE CLEARANCE UTILIZING JOINT SENSITIVITY ANALYSIS

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INTRODUCTION

Tripping and falling, due to decreased toe clearance (TC) during swing, are the greatest sources of accidental injury in children with motor disabilities that affect gait [1]. In children with cerebral palsy (CP), clinicians make predictions about dynamic limb length (DLL) during gait to distinguish between primary and compensatory gait deviations. There is the assumption that poor swing phase clearance is an example of the former while vaulting is an example of the latter dictated by ankle motion [2]. These assumptions are made from analyzing the kinematic data of each joint separately. Moosabhoy et al. developed a method to look directly at joint contributions (JC) as a function of each other in the adult population to assess toe clearance during swing [3]. The aims of this study are to 1) implement the same model in typically developing (TD) children to establish normative data for JC to TC in swing; 2) develop a method to calculate DLL in stance and swing and establish normative data; and 3) to perform a preliminary investigation of the utility in analyzing TC in swing and DLL in stance and swing in children with gait abnormalities.

CLINICAL SIGNIFICANCE

This study demonstrates the utility of quantitative assessment of TC in swing and DLL in stance and swing. Quantifying joint interactions as a function of each other instead of independently provides clinically relevant insight into primary and compensatory gait deviations that can’t be determined from single joint kinematic analysis.

METHODS

This is a retrospective study analyzing kinematic data from twenty children (19 TD, 1 left hemiplegic CP patient). Sensitivity(m/deg) served as a proxy for JC based off the equations described by Moosabhoy et al. [3]. DLL was defined as the distance from hip joint center (HJC) to toe marker divided by the static HJC to floor distance. DLL >1 means the leg is longer than the static standing length. Mean and standard deviation were calculated from the TD children and z-scores were used to calculate statistically significant differences between the child with CP to the TD (z-score ≤-1.96 or ≥1.96).

RESULTS

At minimum TC (66% ±0.029 of gait cycle) in the TD, TC was most sensitive to changes at the hip, then knee, then ankle. In the TD, DLL was the longest at 3 points in the gait cycle: loading response (LR), toe off (TO), and terminal swing (TSW), and 3 points in which it was the shortest: terminal stance (TST), mid-swing (MSW), and initial contact (IC) (Figure 1).

During the peaks, the ankle consistently has a greater percent JC than the knee does in determining DLL. Of the 3 troughs, mid-swing is the only point out of the 3 where both the knee and ankle are working together to shorten DLL. In the child with CP, minimum TC was found at 65% of the gait cycle on the involved side (z-score -0.22). Sensitivity analysis reveals that the patient's ankle is only contributing 0.2% to TC, which differs significantly from the TD (17.1%, z-score -2.567). This decreased ankle contribution is made up for in part by increased ipsilateral knee contribution (49.9% for CP versus 38.5% for TD). The DLL for the subject with CP never appropriately shortens to <1 on the involved side (Figure 1). On the uninvolved side, the child has lengthening of the DLL during stance and does not shorten during MSW (Figure 1). The model identifies a vault occurring on the uninvolved side during the time of minimal TC (66% of the GC) on the involved side (Figure 1). Sensitivity analysis during the vault revealed the knee contributes 49.8% to DLL and the ankle contributes 50.2% to DLL.

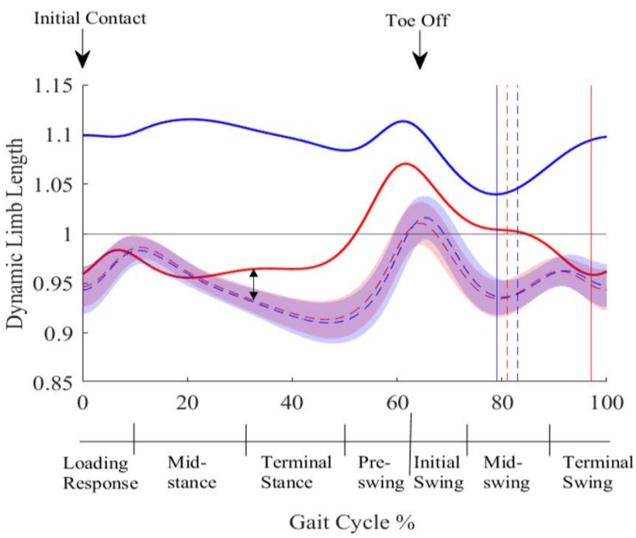

Figure 1: Plot of DLL±1SD for the TD (dashed) and the child with CP (solid). Red corresponds to right and uninvolved side. Blue corresponds to left and involved side. The vertical lines indicate minimum DLL during swing. The double arrow corresponds to the vault on the uninvolved side, which corresponds to the point of minimum TC (66% of the GC) on the involved side.

DISCUSSION

Current understanding of DLL reveals that the leg is longest in stance and shortest in swing [4]. This study redefines DLL to also take into account ankle joint involvement. This method shows DLL to be the greatest during double support, and the shortest during single support in stance. In addition, we can graphically see a compensatory gait deviation (vault) on the uninvolved side. Sensitivity analysis reveals both the knee and the ankle have almost equal contributions in establishing the vault. The ability to look at the individual joints as a function of each other with a sensitivity analysis provides improved insight into the deviations seen in kinematic data.

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DISCLOSURE STATEMENT

Jon R. Davids is a paid consultant for Orthopediatrics Inc. All other authors have no conflicts of interest to disclose.

COMPARING DIFFERENT METHODS OF MEASURING LEG LENGTHS IN PEDIATRIC LEG LENGTH DISRECPANCY

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INTRODUCTION

Anatomical leg length discrepancy (LLD) is defined as the anatomical difference between the lengths of the two lower limbs from the femoral head to the distal tibia [1]. Measurement of LLD has been reported using a variety of techniques, including both radiographic and clinical approaches [2].

At BC Children's Hospital, patients presenting with leg length discrepancy receive a frontal-plane EOS scan (EOS Imaging, Paris, France) in order to visualize the bony anatomy of the lower body and quantify the LLD. The same patients are often referred to The Motion Lab (TML) for a computation gait assessment as part of their standard care. TML utilizes marker-based motion analysis (Qualisys, Göteborg, Sweden) and estimates the locations of the hip joint center (HJC), knee joint centers (KJCs) and ankle joint centers (AJCs) to facilitate kinematic and kinetic analyses. Patients also have their leg length measured from the anterior superior iliac spine to the medial malleolus [2], by a physiotherapist.

The purpose of this study was to compare LLD measurement methods between EOS scans (EOS), motion capture system-based measurement (MoCap), and clinical measurements by a physiotherapist (Physio).

CLINICAL SIGNIFICANCE

This study identifies discrepancies in leg length measurements between three different modalities, and informs clinicians about the potential to utilize MoCap-based LLD measurements in lieu of radiographic based ones. These findings indicate that more research has to be conducted to determine what patient characteristics may result in larger discrepancies between measurement modes of LLD.

METHODS

Children (n=23, age = 12.3±3.1yrs, male n=15) with a LLD of at least 2 cm (measured via EOS) that also underwent a clinical gait analysis at The Motion Lab, Sunny Hill Health Centre at BC Children's Hospital were identified and measurement methods of LLD (EOS, MoCap and Physio) were compared. Analysis included the three measurements done previously as part of routine clinical care. Additionally, EOS scans were used to measure LLD in segments and in its routine clinical care method by two independent reviewers. Inter- and Intraclass Correlation Coefficients (ICC) for all length measurements were determined. 16 patient records contained measurements from all three methods, and 23 patient records contained measurements from only MoCap and EOS. Individual segment measurements from Routine MoCap and EOS segmental re-measurements were compared for both reviewers. The measurements observed for both legs were: AJC to ground, KJC to AJC, HJC to KJC, and overall leg length (HJC to ground). A comparison between methods of measurement was assessed using mean difference and upper and lower limits of agreement (LOA), and visualized by Bland-Altman plots.

RESULTS

ICC values for all EOS measurements were >0.8, indicating high reliability. When measuring the overall leg length, Physio measurements compared fairly well with EOS and MoCap methods, with absolute mean differences of 0.3 and 0.6cm. MoCap and EOS methods exhibited an absolute mean difference of 0.9cm (Table 1). The Bland-Altman plots did not indicate any biases of measurement discrepancy based on the mean magnitude of LLD (Fig. 1). There were four LLD measurements that exhibited differences of >2cm between EOS and MoCap methods, indicating a need to investigate where the discrepancy is manifesting.

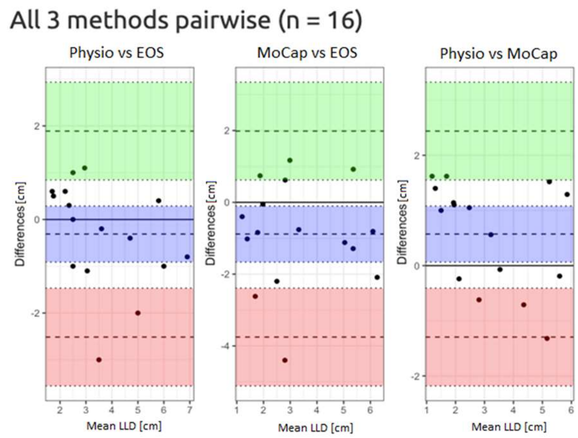

Figure 1 - Bland-Altman plots comparing leg measurement techniques

Table 1: *Summary statistics from Figure 1 Bland-Altman plots.*

Comparison	**Mean Difference [cm]** (95% CI)	**Lower LOA [cm]** (95% CI)	**Upper LOA [cm]** (95% CI)
Physio vs EOS	**-0.3** (-0.9,0.3)	**-2.5** (-3.6, -1.5)	**1.9** (0.8, 2.9)
MoCap vs EOS	**-0.9** (-1.7, -0.1)	**-3.8** (-5.1, -2.4)	**2.0** (0.6, 3.3)
Physio vs MoCap	**0.6** (0.1, 1.1)	**-1.3** (-2.2, -0.4)	**2.4** (1.6, 3.3)

DISCUSSION

While EOS/radiological measurements of LLD are common clinical practice, clinicians should always strive to reduce exposure to radiation, if there are alternate safer, acceptable approaches to obtain the same data. This study has indicated that while mean differences between three measurement methods are all <1cm, there are examples of larger discrepancies that need to be explored and understood before EOS-based, MoCap-based and clinical measurements of LLD can be used interchangeably. This study also indicates the need for continued work with a larger sample size.

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DISCLOSURE STATEMENT

The authors have no conflicts of interest to disclose.

Assessing Gait Variability During Treadmill Walking: is there a Hawthorne effect?

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INTRODUCTION

When evaluating gait in a laboratory or clinical setting, it is necessary to consider the potential effects of the controlled environment. The awareness of being evaluated has been shown to influence gait performance (Hawthorne effect), altering spatiotemporal gait parameters such as step length[1,2], cadence[1], and initial double support time.[3] However, the joint mechanics underlying these modifications are less understood.

Changes in gait during overt observation may not be independent from walking speed. Yet, to our knowledge, no studies have considered controlling for speed when examining differences in gait parameters due to overt vs. covert evaluation. Additionally, gait variability is rarely assessed in relation to the Hawthorne effect, despite its importance in predicting falls.[4]

The purpose of our study was to quantify the Hawthorne effect on gait kinematics by covertly evaluating gait and its variability in healthy young subjects walking on a treadmill.

CLINICAL SIGNIFICANCE

Identifying gait abnormalities in a laboratory or clinical setting is key to developing an appropriate treatment plan. If patients alter their walking when they know they are being observed for that purpose, it can interfere with assessment of their functional status.[5]

METHODS

Healthy adults (18-45 years) with no history of musculoskeletal or neurodegenerative disease were recruited to participate in our IRB-approved study. Passive reflective markers placed on anatomical landmarks according to the full-body plug-in-gait model were tracked with an 8-camera motion capture system (Vicon; Oxford, UK), while participants walked on a motorized treadmill (Motek; Amsterdam, Netherlands) for 4 minutes at a self-selected velocity. Data was collected in two 2-minute segments. Participants were told that the first two minutes was “practice” and was not being recorded, even though it was (“covert” evaluation); then they were notified that the last two minutes were being recorded (“overt” evaluation). We extracted step length, double support time (DST), and sagittal plane range of motion (ROM) of the hip, knee and ankle for each step. Values were then averaged across steps, and averaged values were entered into the analysis. This process was repeated with individual standard deviations to capture stride-to-stride variability.

To date, we have collected data on N=13 participants and anticipate collection on a total of 25 participants by April 2022. For the complete data set, we plan to conduct a repeated-measures MANOVA. Here, we present pilot data from the first 5 participants, focusing on effect sizes. We calculated $d_{\text{repeated measures}}$ (d_{RM}) for each variable. A medium-sized effect ($d_{RM} > 0.50$) would be interpreted as evidence of a potentially meaningful effect that would be expected to show significance in the full sample size.

RESULTS

Among the group means, there was a moderate-to-large sized effect between covert and overt data for only one measure - knee ROM. However, the mean percent change between conditions, while consistent, was quite small (0.79%). Three-of-five variability measures showed medium-to-large effect sizes with a borderline value for DST.

Table 1: Kinematics for N=5 participants under covert vs. overt evaluation (mean± SD). Dark gray boxes correspond to $d_{RM} > 0.50$, and light grey correspond to $0.4 < d_{RM} < 0.50$.

	Group Means			Stride-to-Stride Variability		
	Covert	**Overt**	**d_{RM}**	**Covert**	**Overt**	**d_{RM}**
Step Length (cm)	62.1 ± 11	62.4 ± 11	0.37	1.70 ± 0.6	1.30 ± 0.3	0.87
DST (%)	33.5 ± 2.5	33.4 ± 2.3	0.26	0.75 ± 0.2	0.65 ± 0.2	0.49
Hip ROM (°)	44.0 ± 6.3	44.0 ± 6.7	0.14	1.15 ± 0.3	1.18 ± 0.2	0.14
Knee ROM (°)	62.9 ± 3.4	63.4 ± 3.5	1.83	1.71 ± 0.3	1.48 ± 0.3	0.99
Ankle ROM (°)	32.6 ± 7.9	32.7 ± 7.4	0.06	5.56 ± 3.9	3.87 ± 2.8	0.53

DISCUSSION

Previously reported modifications in spatiotemporal kinematics between covert and overt evaluation may primarily reflect changes in walking speed but may also be driven by subtle changes in knee ROM; if knee motion is a critical factor in clinical evaluation, e.g. in patients with knee osteoarthritis, evaluators may consider covert evaluation. The large effect sizes for stride-to-stride fluctuations highlight differences in gait control even at a constant walking speed. Interestingly, the increased variability seen in fallers could, at least in part, reflect a greater disposition to the Hawthorne effect, although the implications for falling are not clear. Nonetheless, a larger sample size is needed to draw conclusive results on the extent to which the Hawthorne effect impacts gait variability.

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ACKNOWLEDGEMENTS

None.

DISCLOSURE STATEMENT

The authors have no conflicts of interest to disclose.

Should athletes using running-specific leg prostheses be allowed to compete in the Olympic Games?
Dr. Alena Grabowski

In her presentation, Dr. Grabowski will present results from a number of studies that have analyzed the effects of using running-specific leg prostheses on performance during running and sprinting. She will then present data pertaining to a study of an athlete with bilateral transtibial amputations who uses running-specific prostheses and has achieved a 400 m time within Olympic qualifying standards. These data directly address the World Athletics regulation that for an athlete with leg amputations to compete alongside non-amputees in events such as the Olympic games, that athlete must prove on the balance of probabilities that the use of running-specific prostheses does not provide him/her with an advantage compared to an athlete who does not use such prostheses.

Dr. Alena Grabowski earned her Bachelors degree in Kinesiology and PhD in the Department of Integrative Physiology at the University of Colorado Boulder. She then completed a post-doctoral fellowship with the Biomechatronics Group in the Media Lab at the Massachusetts Institute of Technology and earned a career development award within the Department of Veterans Affairs. Dr. Grabowski is currently an Associate Professor in the Department of Integrative Physiology at the University of Colorado Boulder, a Research Healthcare Scientist within the Department of Veterans Affairs Eastern Colorado Healthcare System in Denver, and directs the Applied Biomechanics Lab. Her research program is focused on determining the physiological and biomechanical effects of using mechanical devices such as lower extremity prostheses, exoskeletons, weight support systems, and sports equipment. She studies how these devices affect walking, hopping, running, sprinting, bicycling, and jumping in people with and without physical disabilities such as leg amputations.

Functional Outcomes of Femoral Derotational Osteotomies for Idiopathic Torsional Deformities Using 3D Gait Analysis

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Introduction
Idiopathic torsional deformities (ITD) can present in adolescence with knee pain, patellofemoral instability, running difficulty and difficulty with sports participation [1-2]. This study was performed to evaluate the range of motion and kinematics of patients with ITD following surgery. Secondly, it was to evaluate and compare patient reported outcomes following femoral derotational osteotomy (FDRO) between groups of children with internal and external torsion.

Clinical Significance
Femoral torsion is a common cause of gait disturbance and proper patient selection may help improve outcomes.

Methods
A retrospective review of all fifty-four children with ITD treated with a FDRO at our center between 2000 and 2018 were included. All children completed a pre- and post-operative standardized physical examination and three-dimensional gait analysis (3DGA). Comprehensive gait analysis was also done on 20 typically developing children (TD) as a control group. Patient-reported outcomes results from completed Pediatric Outcomes Data Collection Instrument (PODCI) questionnaires. Statistical analysis was completed using two sample t-tests and Pearson's Correlation.

Results
The internal femoral torsion (IFT) group had 37 patients while the external femoral torsion (EFT) had 17 patients. There were differences in weight with the BMI of 21.1 ± 5.8 kg/m2 for IFT and 32.1 ± 8.6 kg/m2 for EFT. Following FDRO, significant changes to passive external rotation for IFT was 45.4° ± 7.4° and 43.2° ± 8.3° for the EFT while the internal rotation was 43.8° ± 7.5° for IFT and 40.3° ± 12.2° for EFT. The 3DGA revealed a change of 12.2° ± 9.6° for IFT and 12.4° ± 8.0° for EFT. The dynamic mean hip rotation in stance was -1.4° ± 8.3° for IFT and -5.4° ± 8.4° for EFT. The PODCI score significantly improved for the IFT group for transfer/basic mobility, sports/physical function, global functioning and satisfaction with symptoms. The EFT group improved across all domains but only reached significance for satisfaction with symptoms following surgery. Pre-op dynamic hip rotation strongly correlated with pre-op BMI for this study group. ($r=0.57$, $p<0.0001$) (Figure 1).

Discussion
ITD creates significant alteration to normal gait. After surgical correction, both the IFT and the EFT group improved their static ROM within the normative values of our controls. However only the children with IFT walked similar to norms after correction. Children with IFT reported better function than children with EFT. Children with EFT had a significantly higher BMI which

may influence their outcomes. Surgeons treating children with ITD should counsel the children and their families with EFT that their outcomes may not be as favorable as those in children with IFT after surgical correction.

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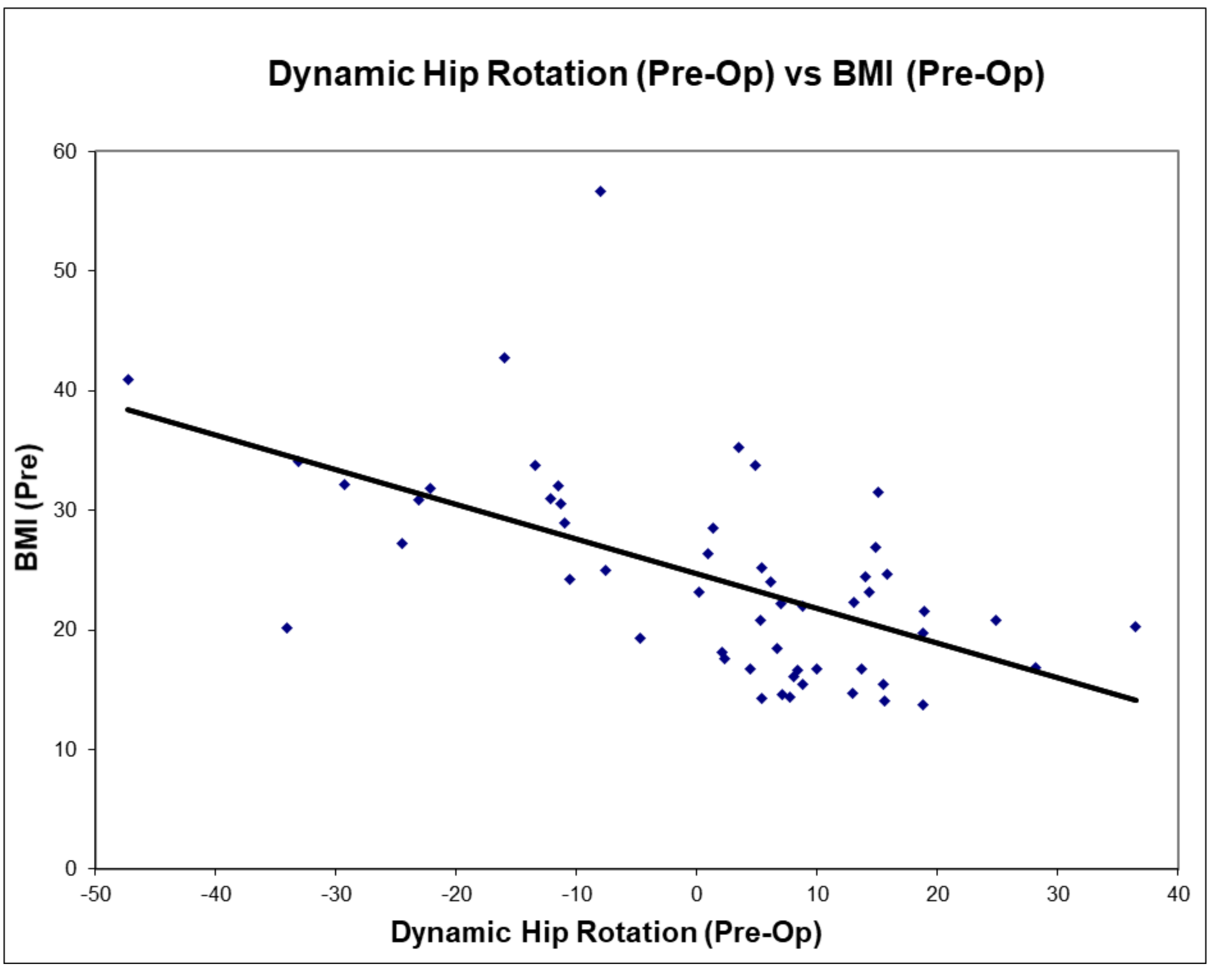
Disclosures:
The authors have no relevant disclosures related to this manuscript.

Figure 1. Pre-operative dynamic hip rotation was strongly correlated to pre-operative BMI for this study group. (r=0.57, p<0.0001).



Long-term Changes in Pelvic Tilt after Hamstring Lengthening during Single Event Multilevel Surgery in Children with Cerebral Palsy

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INTRODUCTION: Flexed knee gait (FKG) is one of the most prevalent gait deviations in children with cerebral palsy (CP). It requires increased energy output and can lead to knee pain. Traditionally, hamstring lengthening (HSL) had been the surgical treatment of choice to correct FKG in children with CP. Satisfactory short term and long-term results have been reported with improved static and dynamic knee extension. However, surgeons have noted unintended consequences after HSL including increased anterior pelvic tilt (APT) which has been linked to increased lumbar lordosis and low back pain and decreasing gait function over time. Some studies have suggested that increased APT may be a short term sequalae of HSL that returns to baseline in the long term (1), while others report that the increase in APT after multilevel surgery (SEMLS) including HSL persists up to 10 years post-operatively (2). The purpose of this study was to assess the short term and long-term APT status post HSL during SEMLS in children with CP.

CLINICAL SIGNFICANCE: When considering surgical correction utilizing HSL for FKG in ambulatory children with CP, surgeons should weigh the likely increased long term APT post operatively with the desired outcome of improved knee extension in stance.

METHODS: This study examined a consecutive sample of ambulatory children with CP, GMFCS I-IV, who underwent HSL during SEMLS and had pre- and two postoperative gait analyses at a pediatric tertiary care center. 44 eligible participants met inclusion criteria (32 male, mean age at surgery 7.2 years, length of short- and long-term follow-up 2.2 and 6.6 years (max 15.9)). 37 out of 42 subjects had medial hamstring lengthening only. Data were extracted retrospectively from the pre-operative (E1) and two post-operative (E2, E3) visits including clinical exam measures and gait kinematics/kinetics. Mean pelvic tilt during the stance phase of gait was compared across visits and as a function of relevant clinical and gait variables using univariate and multivariate linear regression. The relationship of change in pelvic tilt to change in popliteal angle, change in minimum knee flexion in stance and time from surgery to follow-up was examined using bivariate linear regression. The effects of GMFCS level and need for assistance during walking were also examined using multivariable linear regression.

RESULTS: APT increased significantly from E1 to E2 by an average of 4.8° (95% CI: [1.6, 8.0], $p=0.003$). It reverted slightly but not significantly from E2 to E3 by -1.0°, remaining significantly higher by 3.8° at E3 compared with E1 (95% CI: [0.7, 7.0], $p=0.02$) (Fig. 1). Length of follow-up, hip flexor contracture, hip power, hip extension strength, step length, use of assistance and GMFCS level were not related to the mean pelvic tilt or change in pelvic tilt at either short- or long-term follow-up. There was, however, a significant negative relationship of mean pelvic tilt to popliteal angle ($b=-0.22$, 95% CI: [-0.32, -0.11], $p<0.001$) and minimum knee flexion in stance ($b=-0.13$, 95% CI: [-0.20, -0.06], $p<0.001$) which were positively related to each other ($b=0.75$, 95% CI: [0.53, 0.96], $p<0.001$). The change in pelvic tilt was related to the change in knee extension in stance and, to a lesser extent, to the change in popliteal angle (Table 1).

Fig 1: APT by visit, predicted mean (95% CI)

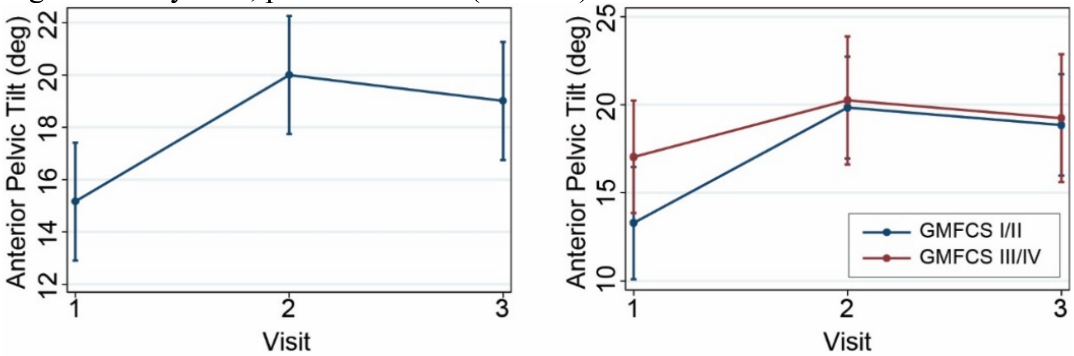

Table 1: Results of linear regression relating change in pelvic tilt to change in popliteal angle and minimum knee flexion in stance

	E1 to E2		E2 to E3		E1 to E3	
	Coef (SE)	P	Coef (SE)	P	Coef (SE)	P
Popliteal angle (°)	-0.13 (0.09)	0.14	-0.15 (0.09)	0.09	-0.27 (0.10)	0.008
Min knee extension in stance (°)	-0.16 (0.07)	0.03	-0.25 (0.07)	0.001	-0.22 (0.06)	0.001

DISCUSSION: Our results indicate that ambulatory children with CP who undergo HSL during SEMLS are likely to have increased APT post-operatively that will persist over time. The increase in APT is associated with the desired outcome of improved knee extension in stance and occurred regardless of GMFCS level, in contrast to previous studies that showed no change in APT after HSL in GMFCS I and II (3). Physicians should weigh the benefits of reducing excessive knee flexion against the risk of increased anterior pelvic tilt for all children with CP undergoing HSL during SEMLS.

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How do psoas lengthening outcomes change after implementation of a machine learning algorithm decision-support tool?

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INTRODUCTION

Hip flexion contracture (HFC) contributes to functional impairment in patients with cerebral palsy (CP) [1]. The primary treatment for HFC and dynamic hip dysfunction is intramuscular psoas lengthening (IMPL), yet, few studies describe specific indications for surgery, and surgical outcome prediction remains difficult. Several studies have proposed various combinations of criteria as an indication for IMPL, including measures of increased pelvic tilt, increased pelvic range of motion in the sagittal plane, or lack of hip extension, with modest improvement in pelvic-hip deviation index (PHiDI) seen in the patients selected to undergo IMPL [2-3]. Recent advancements in machine learning algorithms lead our institution to implement a random forest algorithm (RFA) in 2013 to help with clinical decision-making regarding patient selection for IMPL [4]. This study investigated the impact of this RFA on functional hip outcomes over time and between study groups.

CLINICAL SIGNIFICANCE

Outcome prediction in CP patients undergoing IMPL is difficult; the nonlinear RFA may bolster surgical outcomes, owing to its superior ability to distill a lot of information into one prediction.

METHODS

This was an IRB-approved retrospective study using pre-op and post-op gait lab data from a tertiary children's hospital. Participants had bilateral CP, were 4-17 years old, and underwent 2-7 orthopedic surgeries per limb. The RFA prediction of a good PHiDI outcome (met or failed) was extracted from clinical gait reports. PHiDI measures sagittal plane pelvis and hip gait kinematics (100 = mean typically developing curves). A good outcome was defined as ≥5 point improvement in PHiDI or post-op PHiDI >90. RFA was used to classify limbs into 4 groups [4]:

1) CASES: met, +IMPL
2) CONTROL: met, -IMPL
3) OVER-TREATED: failed, +IMPL
4) OTHER-TREATED: failed, -IMPL

Post-op outcomes were compared between groups and to our outcomes prior to RFA [4].

RESULTS

Ninety-eight people (139 limbs) met the inclusion criteria (CASES: 7, CONTROLS: 87, OVER-TREATED: 11, OTHER-TREATED: 34). GMFCS level (I/II/III/IV) distribution for the 4 groups were 0/2/3/2; 22/36/27/2; 0/7/4/0; 7/8/12/6. CASES and OVER-TREATED were older (mean: 11-12 y), more involved (GMFCS level, gait kinematics), and had greater HFC (Thomas test) (mean: 14-15°). CONTROLS were the youngest (mean: 10 y) and least involved. PHiDI outcomes differed by group (Figure), with cases showing the most improvement, though groups were unmatched at pre-op. For the CASES, CONTROLS, OVER-TREATED, OTHER-TREATED limbs, HFC improved on average 10°, 2°, 6°, and 3°, and good PHiDI outcomes were observed in 43%, 46%, 27%, and 41%, respectively. Overall, good outcomes decreased after the RFA implementation from 58% historically to 43%. Referral practice, CP severity, and proportion of limbs predicted to have a good outcome were unchanged over time. However, historically, 16% had IMPL while 53% did not; after RFA implementation, 5% had IMPL and 63% did not, indicating greater conservatism.

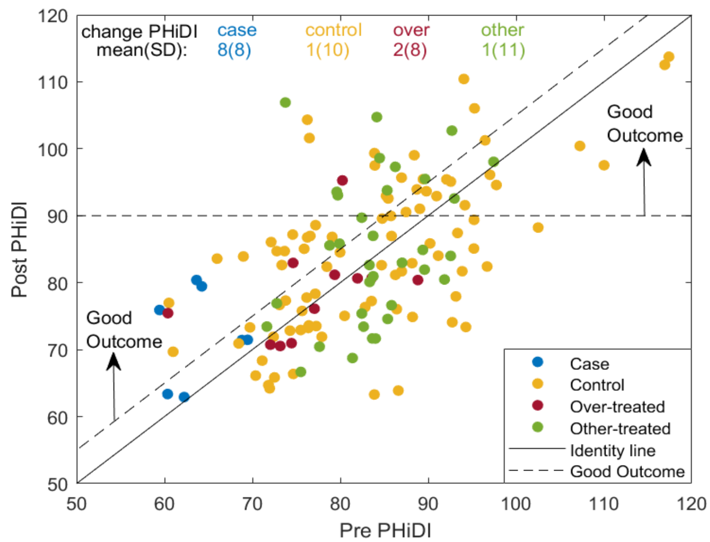

Figure. Scatter plot of pre and post-op PHiDI.

DISCUSSION

Overall, pelvis-hip outcomes worsened since RFA implementation. CASE limbs had the greatest improvement statically and dynamically, though some change may be due to measurement error/regression to the mean given the low pre-op PHiDI. IMPL was often performed on those with HFCs regardless of RFA prediction, indicating deviation from RFA recommendation, likely due to other outcomes being a priority. While these data do not support the use of the RFA to improve PHiDI outcomes for the entire cohort, the +IMPL samples were small and RFA recommendations were not always followed so its efficacy cannot be fully evaluated. PHiDI is one measure; there are other clinical indications and outcomes (e.g. pain, hip subluxation, HFC) that clinicians consider besides the RFA prediction when contemplating an IMPL.

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DISCLOSURES

None.

Is Standing and Transfer Function Improved after Orthopedic Surgery in Youth with CP at GMFCS Levels III/IV?

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INTRODUCTION

Youth with Cerebral Palsy (CP) at Gross Motor Function Classification System (GMFCS) levels III/IV are at risk for losses in standing function during adolescence [1]. Current literature acknowledges multi-level surgery (MLS) as an effective treatment to improve gait, but its effects on standing function are not well documented [2,3]. The objectives of our study were (1) to describe standing function in youth with CP classified at GMFCS levels III/IV, and (2) to evaluate change after MLS.

CLINICAL SIGNIFICANCE

While MLS is regularly completed in children with CP classified GMFCS levels III/IV, literature is limited, focusing on outcomes in this population. Research based outcomes will help to clarify surgical indications and assist in patient and family education regarding realistic post-operative outcomes.

METHODS

In this IRB approved retrospective cohort study, inclusion criteria were youth with CP (GMFCS III/IV) ages 6-20 years who underwent instrumented gait analysis (IGA). A subset who underwent MLS were evaluated for change. Primary outcome measures selected to represent standing function included the Gross Motor Function Measure (GMFM) dimension D, gait velocity, Functional Mobility Scale (FMS), and the Pediatric Outcomes Data Collection Instrument (PODCI). Additional impairment level measures important to standing function included foot pressure, knee extension during stance phase of gait and knee extension PROM. Independent samples Welch's t-tests and chi squared analyses were used to evaluate differences between GMFCS III and IV, and paired samples t-tests and Wilcoxon signed rank tests were used to evaluate change after surgery in each classification group. Statistical significance level was set at $p<0.05$.

RESULTS

A total of 437 children (age 13.7±4.8 years, GMFCS III (81%) and IV (19%)) were included. Ninety-four youth were evaluated for post-operative change 15.3±4.2 months after MLS (mean age 12.1±3.5 years, GMFCS III (74%) and IV (26%)). Youth classified as GMFCS III had higher GMFM-D, gait velocity, PODCI transfer score, and more knee extension compared to youth classified as GMFCS IV ($p<0.05$) (Table 1). After MLS, youth at GMFCS Level III had improved PODCI global scores and Transfer and Basic Mobility subtest scores ($p<0.05$) (Table 1). The GMFCS III group also demonstrated increased knee extension PROM ($p < 0.01$) and improved foot pressure ($p<0.05$) after surgery. Maximum knee extension in stance and heel impulse improved significantly in both groups ($p < 0.01$).

Table 1: Pre vs Post Surgery Comparisons

	GMFCS III (n = 70)		GMFCS IV (n = 24)	
Variables	**Pre-op**	**Post-op (*p*-value)**	**Pre-op**	**Post-op (*p*-value)**
GMFM-D	17 ± 6	19 ± 6 **(0.118)**	8 ± 4	7 ± 3 **(0.568)**
Gait Velocity (cm/s)	52 ± 21	49 ± 25 **(0.265)**	34 ± 16	30 ± 15 **(0.409)**
PODCI				
Transfer and basic mobility	60 ± 17	66 ± 17 **(*p*<0.05*)**	40 ± 17	38 ± 17 **(0.406)**
Global	57 ± 14	61 ± 14 **(*p*<0.05*)**	47 ± 13	45 ± 15 **(0.691)**
Knee extension PROM (°)	-10 ± 12	-7 ± 10 **(*p*<0.01*)**	-14 ± 14	-9 ± 9 **(0.051)**
Max knee extension in stance phase of gait (°)	-27 ± 18	-20 ± 16 **(*p*<0.01*)**	-36 ± 18	-27 ± 18 **(*p*<0.01*)**
Coronal Plane Pressure Index (foot pressure)	30 ± 47	20 ± 44 **(*p*<0.05*)**	30 ± 55	40 ± 52 **(0.324)**
Heel Impulse %	13 ± 18	24±22 **(*p*<0.01*)**	11 ± 15	35 ± 28 **(*p*<0.01*)**
FMS (C/1/2/3/4/5) %				
5 m	27/0/23/35/6/8	10/3/29/44/10/5 **(0.67)**	38/17/46/0/0/0	25/25/46/0/4/0 **(0.9)**
50 m	0/19/53/26/2/0	0/19/50/26/5/0 **(0.9)**	0/46/54/0/0/0	0/75/21/0/4/0 **(0.9)**
500 m	0/84/10/5/2/0	0/76/15/8/2/0 **(0.9)**	0/100/0/0/0/0	0/100/0/0/0/0 **(0.9)**

*Statistical difference between pre- and post-operative values

DISCUSSION

The standing function of youth with CP classified as GMFCS IV was significantly more limited than youth at GMFCS III. After MLS, both groups (III/IV) showed improvement in impairment level outcomes including knee extension and foot position while those at GMFCS III had functional improvement according to PODCI. Orthopedic surgery has the potential to improve standing function in youth with CP classified as GMFCS III however changes in function for youth at GMFCS IV are not well defined. When MLS is intended to improve standing function, a specific protocol of standing and transfer measurements should be adopted for pre-op planning and post-op evaluation.

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PRE-AND POST-SURGERY OUTCOMES FOLLOWING DEROTATIONAL OSTEOTOMY OF THE FEMUR AND/OR TIBIA: PRELIMINARY RESULTS

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INTRODUCTION

About 29% of adolescents have knee pain [1]. One of the underlying causes of this pain is a combination of femoral and tibial torsion, which is often leading to functional limitations and to the cessation of sports activities [2]. Surgical treatment is indicated in patients that do not respond to conservative interventions. The surgical procedure involves femur and/or tibial derotational osteotomies. In order to provide optimal treatment, the effect of these torsional deformities on mobility, function, pain and gait pattern, as well as the effect of surgery on these aspects must be studied.

CLINICAL SIGNIFICANCE

Knowledge gains from this study should lead to the development of best practices for the management of torsional abnormalities.

METHODS

Twenty youths aged 14 to 21 years with a combination of femoral and tibial torsion and candidates for surgery will be recruited at the Shriners Hospitals for Children-Canada. Bone rotational profile will be obtained by a computed tomography scan (CT scan) before surgery. Before and one year after surgery, participants will have a quantitative gait analysis (QGA), a clinical examination (range of motion [ROM] and muscle strength) and they will complete the "Pediatric Outcomes Data Collection Instrument (PODCI)", a patient reported outcomes questionnaire measuring the level of mobility, function, pain and happiness. The Wilcoxon-Signed Rank test was used to assess whether pre-post-surgery changes were statistically significant. The results are reported as: median (95% confidence interval (CI)).

Table 1: Pre-and post-surgery outcomes.

	Measurements	Pre-surgery results	Post-surgery results
Range of motion	Hip internal rotation	70° (55; 80)	50° (45; 59)
	Hip external rotation	25° (20; 35)	45.5° (40; 57)
	Bimalleolar axis	30° (24; 40)	21° (10; 28)
Muscle strength (/5)	Hip flexion	4.8 (4.0, 5.0)	4.5 (4.0, 5.0)
	Hip extension	4.5 (4, 5.0)	5.0 (4.5, 5.0)
	Hip abduction	4.0 (3.5, 5.0)	4.5 (4, 5.0)
	Knee flexion	4.5 (4, 5.0)	4.5 (3.5, 4.5)
	Knee extension	5.0 (5, 5.0)	5.0 (5, 5.0)
PODCI (/100)	Mobility	92.5 (72.0, 99.0)	100 (85.3, 100)
	Sports	52.5 (20.7, 72.2)	76.5 (50.6, 87)
	Pain	37.5 (16.8, 55.2)	71.0 (56.0, 100)
	Global function	70.5 (52.3, 79.7)	88.0 (72.6, 95.3)
	Happiness	77.5 (63.5, 93.3)	82.5 (75.0, 95)

RESULTS

To date, eight participants aged 16.6 years (95% CI: 14.9 years, 20.9 years) have completed the study. Baseline CT scan results demonstrate increased femoral anteversion (31° [95% CI: 15°, 46°]; norms: 14°

±7.6 °) and increased tibial torsion (47° [95% CI: 35°, 72°]; norms: 31.9° ±6.7 °). The surgery consisted of a derotational osteotomy of the femurs in 14 legs (25° external [95% CI: 18 °, 30°]) and the tibia in 11 legs (20° internal [95% CI: 13°, 25°]). The pre-and post-surgery outcomes can be found in table 1 and QGA data are presented in figure 1.

At 13.4 months (95% CI: 12.2 months, 15.8 months) post-surgery, statistically significant improvements were observed in the following domains of the PODCI: sports and physical function with 16 points (95% CI:12.7, 39.3; p=0.01), pain and comfort with 29 points (95% CI:17, 67; p=0.01) and for the global function with 15 points (5.75, 31.2; p=0.008). These improvements might be explained by the normalization of the physical examination and the QGA. For the clinical assessment, significant changes were observed for the hip internal rotation ROM with a reduction of 20° (95% CI: -30°, -6°; p=0.001), the hip external rotation ROM with an increase of 23.5° (95% CI: 13 °, 30 °; p=0.001) and for the bimalleolar axis with a decrease of 15° (95% CI: -17°, -4°; p=0.004). For the QGA, a significant decrease of 14° for the mean hip rotation during stance (95% CI: -13, -15 °; *p* <0.001) and an increase of 12.3° (95% CI: 11.9 °, 12.6 °; *p* <0.001) for the mean knee rotation during stance. No statistically significant changes were observed for muscle strength.

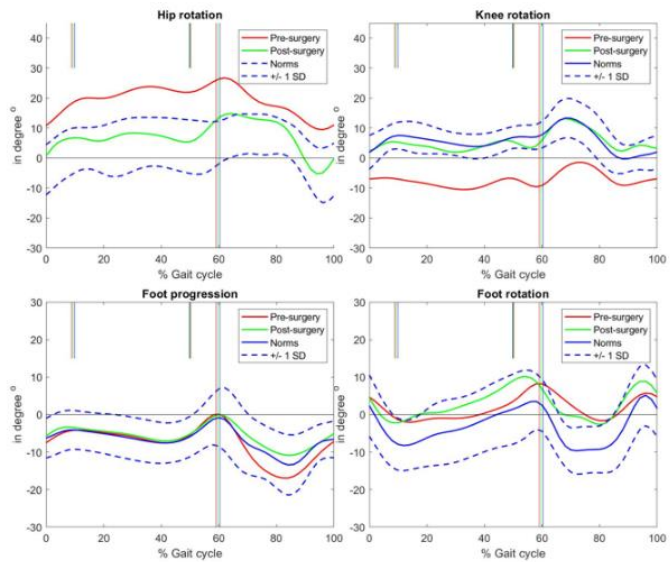

Figure 1. Pre-and post-kinematic data in the transverse plane

DISCUSSION

Preliminary results suggest that participants seem to benefit from this extensive surgical procedure as shown with the improvement in the different PODCI domains, in their gait pattern and in their ROM. However, it is unclear what are the consequences of the surgery in the long term and longer studies are needed.

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DISCLOSURE STATEMENT

Mitchell Bernstein MD is a consultant for Smith and Nephew, Nuvasive and Orthofix. The other authors have no conflict of interest to declare.

Age-associated changes in vertical support strategy with increasing walking speed

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INTRODUCTION

Each year approximately one-third of elderly adults over the age 65 fall at least once. Elderly adults commonly walk with decreased limb support force as characterized by vertical ground reaction forces (VGRFs) [1]. In addition, aging may cause a distal to proximal redistribution of joint moment contributions to lower limb force production during gait [2]. For example, in younger adults, the knee extension moment is the main contributor to the 1st peak of VGRF during weight acceptance phase of walking. However, when gait velocity was matched, older adults showed greater hip extension joint moment/power with reduced knee extension and ankle plantarflexion moment/power compared to younger adults. Compared to the sagittal plane, frontal plane joint moments that contribute to VGRF have received less attention. According to simulation studies, gluteus medius activity contributes to the 1st peak of VGRF and supports the bodyweight in the early weight acceptance phase [3]. However, the contributions of changes in lower extremity joint moments to changes in VGRF during speed modulation have not been documented. Therefore, the primary purpose of this study was to assess how aging affects the relationships between changes in joint moments and changes in VGRF across walking speeds.

CLINICAL SIGNIFICANCE

Decreased limb support forces has been associated with slower walking speed and increased risk of falling in elderly adults. Understanding how aging affects the lower extremities joint moments contribution to limb support force generation will provide a benchmark for assessing a potential gait deficit in this population that may provide insight for rehabilitation programs aimed at reducing the rate of falling with aging.

METHODS

Twenty-four young and eighteen healthy elderly adults from the previous open-source dataset were analyzed [4]. All subjects walked on a split-belt instrumented treadmill (FIT; Bertec, Columbus, OH, USA). This study analysed the preferred walking speed (PWS) and 30% faster (FS) than PWS trials. Kinematic and kinetic data were collected at 150 Hz and 300 Hz, respectively, using 26 reflective markers on the pelvis and lower-extremity segments with a 12-camera system (Motion Analysis, Santa Rosa, CA, USA). Kinetic and kinematic variables were calculated in Visual3D (C-motion Inc., Germantown, MD, USA). Stepwise linear regression was used to determine the primary predictors of the changes in each peak across speeds. The significance level was set at 0.05. All statistical analysis were performed in R.

RESULTS

In younger adults, changes in knee extension moment accounted for 49% of the variances of changes in the 1st peak of VGRF with speed increase ($\beta = 7.39$, $p < 0.001$). In elderly subjects, changes in hip extension moment ($\beta = 8.44$, $p < 0.001$) accounted for 68% of the variances in the 1st peak of VGRF during speed increase (Fig.1-A). In younger adults, the change in ankle plantarflexion moment (β = 4.98, p = 0.005) and the change in hip abduction moment (β = 3.97, p = 0.022) accounted for 53 % of the variances in changes in the 2nd peak of VGRF. In elderly subjects, the final model of change in the 2nd peak of VGRF included only change in ankle plantarflexion moment as a predictor (β = 8.02, p < 0.001), which accounted for 71 % of the variances of change in the 2nd peak of VGRF (Fig.1-B).

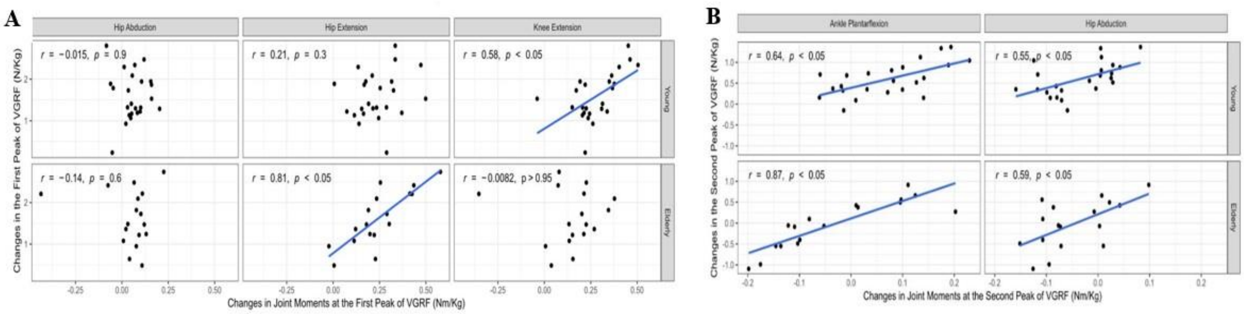

Figure 1. A: Relationship between changes in the 1st peak of VGRF and changes in lower limb joint moments from PWS to FS in young and elderly subjects. **B:** Relationship between changes in the 2nd peak of VGRF and changes in lower limb joint moments from PWS to FS in young and elderly subjects.

DISCUSSION

Younger adults mainly increased knee extension moment to increase the 1st peak of VGRF during speed modulation. In contrast, elderly subjects primarily modulate their hip extension moment to increase the 1st peak of VGRF with increasing speed. These results may indicate that elderly adults have reduced capacity to increase knee extension moment to modulate VGRF and is consistent with prior research that reported that elderly adults increased their hip extension moment when asked to walk faster [5]. In younger subjects, the changes in ankle plantarflexion and hip abduction moments were the main contributors to changes in VGRF during speed modulation. In contrast, the changes in ankle plantarflexion moment in elderly was the main contributor to changes in VGRF during speed modulation. Consistent with previous studies ankle plantarflexion moment in elderly subjects was lower than younger adults at both speeds. Additionally, most of the elderly subjects did not increase their ankle plantarflexion moment and hip abduction moments when walking faster. Our results indicated that the limited capacity to increase ankle plantarflexion and hip abduction moments likely reduced the ability to increase VGRF in elderly adults.

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DISCLOSURE STATEMENT

None.

QUANTIFYING POSTURAL CONTROL STRATEGY RELIANCE USING FORCE INTERSECTION POINT ANALYSIS AMONG SYMPTOMATIC DEGENERATIVE SPINAL PATHOLOGIES

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INTRODUCTION

Postural balance deficits are a common symptom among degenerative spinal pathologies (DSP). There has been growing clinical interest in the use of objective assessment of balance among DSP surgery patients to indicate both severity of preoperative symptoms and postoperative outcomes [1, 2]. To date, most methods of assessing DSP patient balance have focused on postural sway measures which have inherent limitations in interpretation of underlying drivers of postural instability and related stabilizing actions. Recently, a novel method of quantifying postural control reliance was developed called force intersection point (FIP) analysis which provides insight into body stabilization modes [3]. The relevance of FIP analysis to DSP patients has yet to be determined and may provide new insight into the impact of surgical treatments on patient's postural control strategy.

CLINICAL SIGNIFICANCE

Quantification of balance strategy reliance using FIP analysis may provide unique insight into balance deficits among DSP patients and might serve as a clinical screening tool to aid in optimizing treatment timing decisions and assessing treatment outcomes.

METHODS

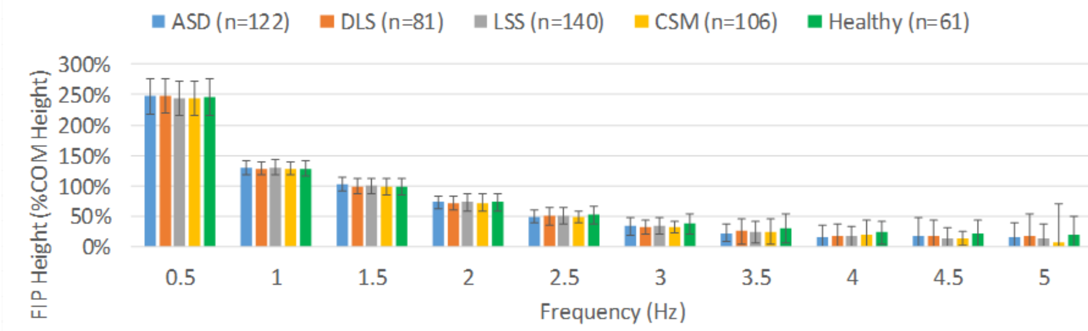
A retrospective review of symptomatic DSP patient balance data was conducted at a single institution. DSP patients were included from four cohorts: adult spinal deformity (ASD), degenerative lumbar spondylolisthesis (DLS), lumbar spinal stenosis (LSS), and cervical spondylotic myelopathy (CSM). Healthy (H) control data was also included for comparison.

Data was collected from a force plate sampled at 2kHz during 60s standing tests with subject's eyes open and arms at their side. Ground reaction force (GRF) and center of pressure (COP) data were low-pass filtered at 100Hz. Sagittal-plane GRF angle and COP location were then each band-pass filtered using 10 bands of 0.5Hz width from 0.5–5Hz. Within each band the FIP was calculated and expressed as % height of center of mass (COM).

RESULTS

A total of 449 DSP patients and 61 H subjects were included (avg±1std): ASD: 85F/36M, 61±16yr, 1.6±0.1m, 77±21kg; DLS: 51F/31M, 60±15yr, 1.7±0.1m, 81±16kg; LSS: 40F/100M, 54±17yr, 1.7±0.1m, 92±22kg; CSM: 58F/48M, 62±10yr, 1.7±0.1m, 82±19kg; H: 37F/24M, 40±15yr, 1.7±0.1m, 72±15kg. DSP patients reported clinically significant pain on visual analog scales (VAS: neck (CSM)=4.8, back>6.0, leg>4.0) and disability on the Oswestry

Disability Index (ODI>40) or Neck Disability Index (NDI (CSM)>40). Cohort averages for FIP height by frequency are shown in the Figure.



Difference	0.5Hz	1Hz	1.5Hz	2Hz	2.5Hz	3Hz	3.5Hz	4Hz	4.5Hz	5Hz
ASD-H	1.5%	2.0%	3.4%	-0.3%	-3.6%	-4.3%	-8.4%*	-6.5%*	-4.4%	-4.5%
DLS-H	1.7%	0.5%	0.2%	-1.6%	-2.6%	-5.0%*	-5.9%	-4.6%	-5.0%	-1.9%
LSS-H	-1.6%	1.9%	1.3%	-0.6%	-2.2%	-3.6%	-7.1%*	-5.9%*	-8.1%*	-5.3%
CSM-H	-1.6%	0.5%	-0.3%	-1.1%	-4.1%*	-5.1%*	-6.8%	-3.9%	-8.0%*	-12.8%

Figure: Average FIP height by study group (top) and % difference relative to H (bottom). Error bars indicate one standard deviation and * indicates significant at $p<0.05$.

ASD FIP was lower at 3.5Hz (23%, p=0.003) and 4.0Hz (17%, p=0.029) compared to H (32% and 24%). DLS FIP was lower at 3.0Hz (33%, p=0.036) than H (38%). LSS FIP was lower at 3.5Hz (25%, 0.018), 4.0Hz (18%, p=0.030), and 4.5Hz (14%, p=0.005) than H (32%, 24%, 22% respectively). CSM FIP was lower at 2.5Hz (49%, p=0.027), 3.0Hz (33%, p=0.011), and 4.5Hz (14%, p=0.003) than H (53%, 38%, and 22% respectively).

DISCUSSION

FIP analysis revealed slight differences in DSP patient balance strategy reliance compared to H which may shed light on the impact of DSP on balance control. In particular, although not significant, there was a nearly consistent trend across symptomatic DSP cohorts in elevated FIP height below 1.5Hz and reduced FIP height above 1.5Hz. Further investigation is warranted to better understand FIP behavior within each cohort with specific etiologies in mind. Quantifying balance strategy with FIP analysis may hold value as a future screening tool for DSP and in assessing postoperative outcomes.

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DISCLOSURE STATEMENT

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LOWER EXTREMITY JOINT STIFFNESS DURING RUNNING IN ADOLESCENTS WITH AUTISM SPECTRUM DISORDER

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INTRODUCTION

Running is the most common form of physical activity for girls and the second most common form for boys aged 12 to 15 years [1]. This is consistent for autistic youth, who reportedly enjoy solitary activities, such as running, more than team-based sports [2-4]. Autism spectrum disorder can be linked to a wide variety of qualities influencing motor skills, social communication, and behavior characteristics that may differ from non-autistic peers. Autistic youth have reported elevated levels of fear for sustaining injury, being bullied, and fear of exclusion within their physical education classes compared to their non-autistic peers [4], emphasizing the importance of investigating solitary physical activities within this population.

It is also important to note late adolescents and teens undergo rapid skeletal growth changes when the epiphyseal plates weaken and skeletal muscle strength increases, leaving individuals more vulnerable to injury risk. Inadequate joint stiffness is one of several factors that may influence injury risk [5]. The purpose of this study was to examine ankle and knee joint stiffness in autistic adolescents and teenagers and non-autistic matched controls at self-selected and matched running speeds.

CLINICAL SIGNIFICANCE

Investigating loading and joint stiffness of the lower extremity in persons with autism spectrum disorder during running may provide points of emphasis for therapeutic and rehabilitation interventions.

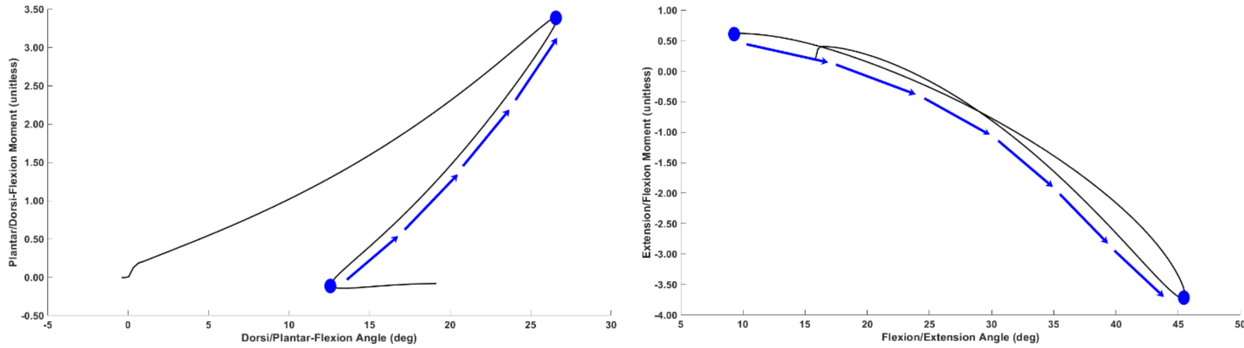
METHODS

Twenty-two persons with a confirmed autism spectrum disorder (ASD) diagnosis (15 males, aged 14±2 years, BMI 22.24±5.76 kg/m^2) and seventeen age, sex, and body mass index healthy controls (CON) (12 males, aged 13±1 years, BMI 16.54±3.58 kg/m^2) were prospectively enrolled into an IRB-approved study.

Kinematic and force data were collected using a 10 camera VICON motion capture system sampled at 200 Hz simultaneously with ground reaction forces using three in-line Bertec force plates at a frequency of 2000 Hz. Participants were asked to complete a series of over-ground running trials at their self-selected speed and a standardized speed of 3.0 m/s. The average self-selected speed for the ASD group was 2.52m/s; the CON group were asked to match (within ±5%) the self-selected speed of their match with ASD. Five successful force plate strikes of each leg (total=10) for each participant were analyzed in Visual 3D (Version 6, C-Motion, Inc.). Marker and force data were low pass filtered with a cutoff frequency of 10 Hz.

Joint stiffness (unitless) was calculated as the quotient of the change in joint moment (unitless; normalized to body mass * leg length) and the change in joint angle (radians) during the energy absorption period of stance (Figure 1). Statistical analyses were performed in SPSS (version 27, IBM Corp.). Stiffness and changes in joint moments (a component of stiffness) were analyzed using 2 (group) x 2 (speed) analyses of variance (ANOVA).

Figure 1. Examples of ankle (left) and knee (right) joint stiffness waveforms.



RESULTS

There were no significant interactions between groups and speeds ($p>0.05$) for any joint variable. Persons with ASD had reduced knee and ankle joint stiffness (all $p<0.020$; Table 1). Running at the 3.0m/s standardized speed resulted in increased knee joint stiffness compared to self-selected ($p=0.010$). Persons with ASD had reduced changes in knee and ankle moments ($p<0.003$; Table 1). Running at the 3.0m/s standardized speed increased knee moments over self-selected speed ($p=0.004$).

Table 1. Ensemble joint stiffnesses and change in joint moments (mean±SD).

	Joint Stiffness				Joint Moments			
	ASD SS	ASD 3.0	CON SS	CON 3.0	ASD SS	ASD 3.0	CON SS	CON 3.0
Knee	5.84±0.86	6.44±0.80	6.80±1.05	6.97±1.32	2.47±0.35	2.67±0.38	2.80±0.33	2.90±0.42
Ankle	7.49±1.15	7.33±1.43	8.88±2.35	8.77±2.28	2.28±0.35	2.38±0.38	2.81±0.42	2.82±0.43

DISCUSSION

Joint stiffness during running has typically been viewed as "greater stiffness indicates an increase in injury risk" [6]. Given the reports from current literature regarding joint stiffness during walking in those with ASD, we expected to also find a similar increase in joint stiffness during running [7]. However, those with ASD had reduced stiffness, which was due to their reduced joint moments. Decreased joint stiffness by those with ASD could be indicative of a less efficient running style whereby the elastic recoil is not being optimally utilized by the knee and ankle musculature. However, we cannot ignore the implications of prior training on running mechanics. We did not ascertain the physical activity and education history of the participants; however, we did track physical activity engagement with our previous work, finding no differences between groups in low and moderate to vigorous physical activity [8].

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DISCLOSURE STATEMENT

No conflicts of interest to disclosure

FEMALES WITH ACETABULAR DYSPLASIA AND LABRAL TEARS WALK WITH DECREASED HIP EXTENSION ACROSS TIME SERIES

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INTRODUCTION

Hip osteoarthritis (OA) is a chronic musculoskeletal condition that causes pain and functional limitations in daily life. One of the most prominent gait features of individuals with hip OA is decreased hip extension. Females with mild to moderate hip OA and limited hip extension during gait have reported worsening hip pain [1]. Acetabular dysplasia, a hip morphology with decreased acetabular coverage over the femoral head, increases the risk of developing hip OA [2]. Furthermore, acetabular labral tears are a precursor to hip OA and are more common in females [3]. Individuals with hip dysplasia also walk with greater anterior pelvic tilt, which also affects hip angles [4]. Therefore, we studied the range of the gait cycle over which females with acetabular dysplasia and labral tears walked in less hip extension than females without pain while observing contributions from the pelvis.

CLINICAL SIGNIFICANCE

Monitoring movement patterns and applying rehabilitation techniques prior to arthritic changes are crucial in delaying or avoiding hip OA.

METHODS

Adolescent and adult females diagnosed with dysplasia or labral tears (hip pain group) and healthy controls without hip pain were recruited. Hip pain participants were on average 26.7 years, 68.9 kg, and 1.67 m. Control group participants were on average 23.8 years old, 61.3 kg in mass, and 1.64 m in height. All participants provided consent. Kinematic data were collected while walking on a treadmill at two speeds: preferred and fast (25% faster than preferred). Peak hip extension and pelvis angles were calculated at each walking speed. We conducted a 2-sample unequal variance t-test to observe group differences between peak hip extension. To explore beyond discrete peak angles, we used statistical parametric mapping (SPM) [5] combined with 2 sample t-tests to analyze the one-dimensional time series data.

RESULTS

Kinematic data from the hip pain (N=39) and without hip pain (N=35) groups were compared. During fast speed, peak hip extension of the hip pain group was significantly less than that of the control group ($p=.029$) with a difference of 3.9°. In fast walking, the time series analysis depicted less hip extension in the hip pain group (Fig. 1C) from 43.1% to 52.7% of the gait cycle ($p=.046$). There were no significant differences in hip extension during preferred walking from discrete or time series analyses. The anterior pelvic tilt was greater for the hip pain group than the control group (Fig. 1D) during the entire gait cycle in fast walking speed ($p=.002$) and preferred walking speed ($p=.003$).

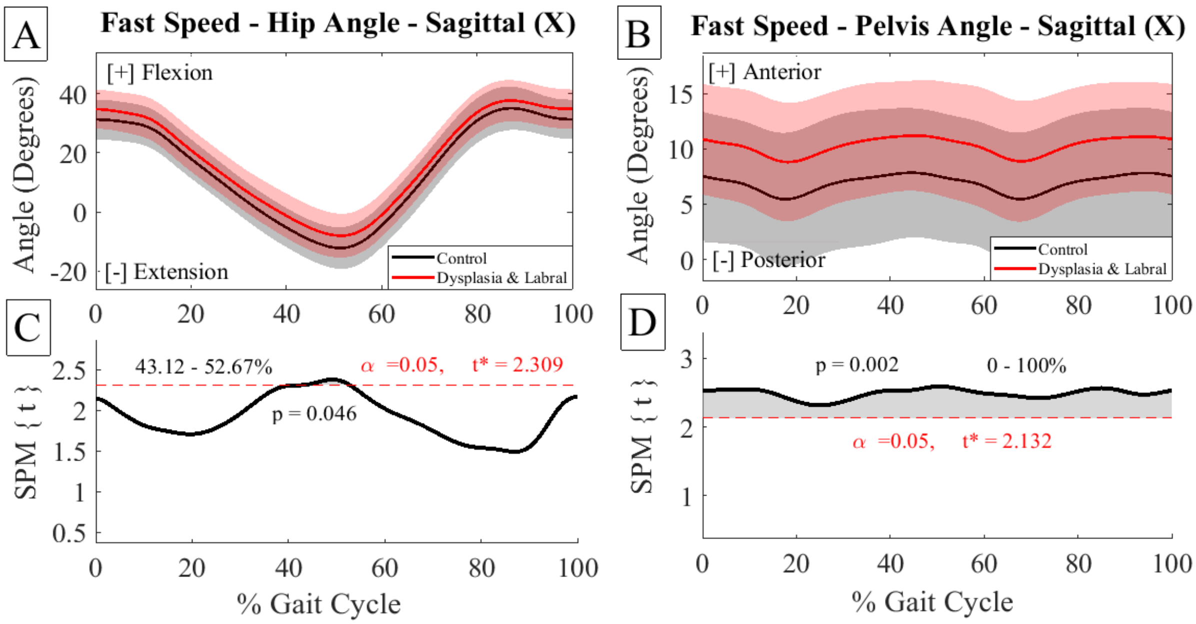

Figure 1. Average sagittal hip angles (A) and pelvic angles (B) for the 2 groups during fast walking. SPM one-tailed t-test for sagittal hip angle (C) and pelvic angle (D) comparisons.

DISCUSSION

This study shows that decreased peak hip extension, common in females with hip OA, exists for individuals with acetabular dysplasia and labral tears, preceding arthritic changes. The time series analysis further confirms that decreased hip extension persists beyond the peaks seen with discrete data. Despite consistently increased anterior pelvic tilt (Fig. 1B) and a more flexed hip position throughout the gait cycle (Fig. 1A), the difference was only significant during terminal stance and early pre-swing in fast walking. The significantly increased anterior pelvic tilt persisted in preferred walking as well, but the observed differences in hip extension were not significant at this speed.

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DISCLOSURE STATEMENT

The authors have no conflicts of interest to disclose.

INDEPENDENTLY AMBULATORY CHILDREN WITH SPINA BIFIDA EXPERIENCE NEAR-TYPICAL JOINT LOADS DURING WALKING

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INTRODUCTION

Spina bifida is a neurological birth defect that often results in lower limb muscle weakness. Children with spina bifida have diminished motor abilities and engage in less activity than peers with typical development [1]. This may increase risk of osteoporosis and low-energy fracture.

Bone strength increases with ambulation and function in spina bifida, and individuals with high function can experience typical bone health [2]. Altered gait kinematics may affect the dynamic mechanical loads important for bone accrual and maintenance [3]. The purpose of this study is to investigate whether joint loads during walking differ between independently ambulatory children with spina bifida and children with typical development. We hypothesized that knee and ankle loads would not differ between groups, given the high function observed in both.

CLINICAL SIGNIFICANCE

Our results suggest that during walking, independently ambulatory children with spina bifida undergo near-typical knee and ankle loads, which influence bone accrual and contribute to near-typical bone health. Independent ambulation should be promoted to preserve bone strength.

METHODS

Data for this study were collected at Children's Hospital Los Angeles [2]. Independent ambulators with spina bifida (10.4 ± 2.7 years old, 8 female / 8 male) and children with typical development (n=16, age- and sex-matched) walked overground at self-selected speeds, barefoot and unassisted. Ground reaction forces and marker data were acquired through floor-embedded force plates (AMTI, Watertown, MA) and a motion capture system (Vicon, Oxford, UK). Depending on data available, 1-5 gait cycles were used per participant. Quantitative computed tomography and custom scripts were used to estimate 8 properties of the tibia. A StepWatch (Modus, Edmonds, WA) measured daily steps of 14 participants with spina bifida over 1-9 days.

We estimated joint kinematics and kinetics with a musculoskeletal model [4] in OpenSim 4.1 [5]. Using inverse dynamics and static optimization, we estimated lower limb muscle forces and shear and compressive tibia loads at the knee and ankle. For each participant with spina bifida, we analyzed the more affected leg. If both legs were equally affected, and for participants with typical development, we analyzed the leg with more gait cycles. We averaged joint loads across cycles and used Statistical Parametric Mapping (SPM) t-tests to compare average waveforms between the groups with spina bifida and with typical development. We also used Welch's t-test to compare peak loads, walking speed, and bone properties between groups.

RESULTS

No statistically significant differences in average shear and compressive loads at the knee and ankle were observed between the two groups (Fig. 1). Differences in mean peak joint loads were not statistically significant ($p>0.15$).

Except for cancellous bone density in the tibia's proximal metaphysis (p=0.001), bone properties did not differ (p>0.25) between groups. Walking speed was slower (p<0.001) in the group with spina bifida (1.1 ± 0.1 m/s) than in the group with typical development (1.3 ± 0.1 m/s). Average daily steps in the group with spina bifida (9656 ± 3095) were similar to those of an external group with typical development (9589 ± 3322) that underwent the same protocol [6].

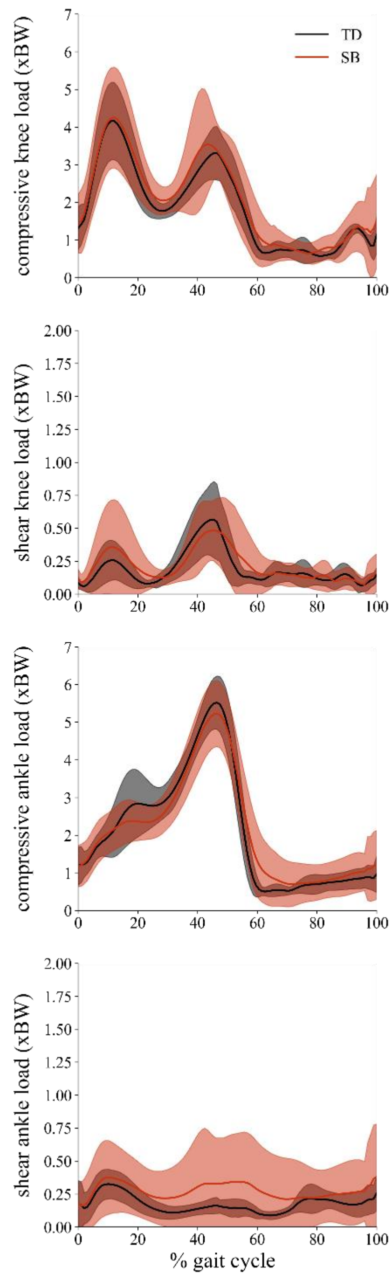

Figure 1. Mean ± SD joint loads across the gait cycle for the groups with typical development (TD, black) and spina bifida (SB, red). No significant differences (p>0.05) were identified by SPM t-tests.

DISCUSSION

Despite slower walking speeds and weaker lower-limb muscles, the group with spina bifida exhibited near-typical bone strength and knee and ankle loads during walking. Greater load variation was observed in the group with spina bifida.

Numbers of steps per day were similar between the groups with typical development and with spina bifida. These typical load magnitudes and cycles observed in the group with spina bifida, in conjunction with typical bone health, support that dynamic mechanical loading is important for bone maintenance.

This study included a small sample of young, high-function, independently ambulatory individuals walking in-lab. Future studies should focus on relationships between loading and bone health in a more diverse population of individuals, including those post-puberty and with lower function. Further study of loading is also needed for activities in free-living settings.

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DISCLOSURE STATEMENT

The authors have no conflicts of interest to disclose.

DEVELOPING A GAIT SUMMARY METRIC USING AN AUTOENCODER

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INTRODUCTION

Individuals with cerebral palsy (CP) often present with complex gait dysfunction affecting multiple lower extremity segments in multiple planes of movement. Treatment is often facilitated by employing instrumented gait analysis which produces an abundance of quantitative features including temporal-spatial parameters, kinematics, and kinetics. The Gait Deviation Index (GDI, [1]) is a summary metric of gait that represents a linear combination of kinematic features. The GDI has been used to measure the severity of gait dysfunction and evaluate the effectiveness of various interventions on improving gait quality. A larger array of gait features is available which has a variety of nonlinear relationships that can be integrated for a more complete summary metric of gait. The current work introduces a new model that builds from the GDI by not only including temporal-spatial parameters but also evaluates linear and non-linear relations among gait features. Therefore, we propose a deep learning autoencoder model that will learn from gait features including kinematic and temporal-spatial parameters and compresses the data into a single score capable of assessing gait quality and the effectiveness of interventions.

CLINICAL SIGNIFICANCE

Having a single score to represent the quality of human gait can be helpful for clinicians when identifying gait dysfunction, evaluating progress, and treatment outcomes.

METHODS

Instrumented gait analysis data from Shriners's Motion Analysis Centers (MAC) - Chicago, Texas, Philadelphia, and Northern California were queried for subjects under the age of 18 diagnosed with CP (CP Group). Data queried included gait kinematics (64 features), kinetics (43 features), temporal-spatial parameters (7 features), and Gross Motor Function Classification System (GMFCS) level. Additionally, a Control Group comprised of typically developing children was also queried. The autoencoder model is a dense, fully connected, neural network with a dropout layer between the input and bottleneck layer that was developed in the Python programming language using the TensorFlow package. The data was preprocessed first by removing samples which were missing any features, then a min-max normalization was applied so that features with large values were not favored. Then, a subject-wise 10-fold cross validation was used to avoid overfitting and trained the tuned autoencoder model to capture the statistical regularities in the data. Model performance was evaluated using explained variance. The autoencoder (AE) score was then calculated by using the mean and standard deviation of the Control Group to indicate the quality of their gait. Feature importance analysis was then performed to identify important features of the autoencoder score. We also performed cross-validation across the different locations to validate the process. Finally, we performed Pearson correlation analysis and built prediction models to evaluate the AE.

RESULTS

Data from 2107 subjects in the CP Group (GMFCS I: 25%, GMFCS II: 44%, GMFCS III: 21%, GMFCS IV: 3%, NR: 6%) and 144 subjects in the Control Group were included for the study. To include subjects who required an assistive device (GMFCS III, IV), only temporal-spatial and kinematic features (71 features in total) were used for the autoencoder model. The cross-validation results show that the autoencoder results were consistent across sites (Table 1). Walking speed was found to be the most important feature for the AE (Figure 1). The AE was found to have a higher correlation to GMFCS level than GDI. (AE: 0.56, $p<0.001$; GDI: 0.53, $p<0.001$). The model using GDI to predict GMFCS level had a root mean squared error (RMSE) of 0.80, while the model using the AE had a RMSE of 0.78.

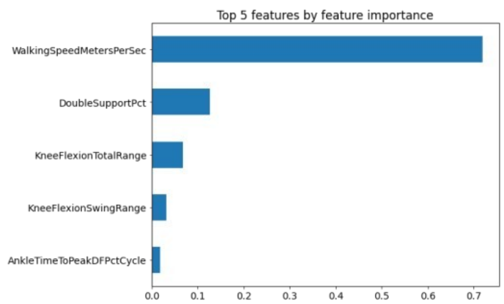

Figure 1. Feature importance analysis shows that Walking Speed has the highest importance for the autoencoder score.

Table 1: Explained variance (%) captured by the cross-validation (CV) autoencoder model across each site using a model developed with Chicago data.

Model	CHI	TX	PHL	NorCA
CHI	88.92 ±2.58	89.26 ±1.86	89.41 ±1.79	86.17 ±2.09

SUMMARY

Gait has many interconnected characteristics. Using a deep learning model, we were able to capture relationships among the measured gait parameters. The results of this study show that the autoencoder generated from the model predicted GMFCS levels. Cross validation demonstrated that the AE model was generalizable across multiple sites. To ensure the score generated through the proposed autoencoder reflected gait quality, additional studies evaluating gait progression over time, or the effects of intervention are warranted. We are working on a version of the autoencoder that also includes kinetic features.

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DISCLOSURE STATEMENT

Jon Davids, MD is a consultant for Ortho Pediatrics; All other authors have no conflicts of interest to disclose.

QUANTIFYING CONCURRENT VALIDITY OF GAIT KINEMATICS PROVIDED BY XSENS MVN IMU SYSTEM

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INTRODUCTION

In 3D human motion study, inertial measurement unit (IMU) system offers purported benefits over optoelectronic measurement system (OMS) in terms of portability, cost-effectiveness, and real-time analysis. Although the real-time analysis of functional daily life movements outside controlled laboratory environment introduced a variety of potential confounding variables to the study, it gave important clinical information a therapist may use to better assist their patient in therapy without needing access to a specialized biomechanical laboratory. During last four decades, OMSs have been providing comprehensive and objective musculoskeletal data for clinical assessment and thus regarded as 'reference of measurement in 3D human motion' by the clinicians [1]. IMSs being a potential alternative to OMSs, they need to be verified concurrently with the OMSs before applying to use IMSs in the clinical environment.

We conducted the concurrent validity study of Xsens system (IMU) in comparison to Vicon system (OMS) for lower extremity movements during walking and jogging in healthy individuals. The aim of this study was to determine the degree of similarity of kinematic data collected between the Xsens and Vicon systems in 3D human motions.

CLINICAL SIGNIFICANCE

The result showed that the Xsens did not produce accurate kinematic data when compared with the VICON system. The differences in angular displacement of lower extremities in gait and jog were larger than 5 degrees. However, the Xsens device was effective at capturing movement and producing appropriate waveforms for the ankle, hip, and knee during normal gait. For this reason, we determined Xsens has the potential to be used in a clinical setting when no other resources were available. We recommend that Xsens system not be utilized to clinically measure exact joint angles and was not able to replace the current VICON system. However, the Xsens could be appropriately utilized to measure the change in large motion or gait abnormalities in settings where OMS was not practical or accessible. The portability and ease of use of the Xsens allowed clinicians to perform real-time motion analysis without the need for a specialized laboratory.

METHODS

Ten healthy adults (3 females, 7 males; 24.9 ± 1.6 yrs.; 76.3 ± 19.8 kg) were recruited and after providing consent served as participants in the study. Seventeen Xsens sensors were attached halfway between each body segment according to Xsens manual. In Vicon system, anatomical marker placement was done by utilizing a Plug-in Gait (PIG) model. Angular displacements of the lower extremity were concurrently measured by both systems during walking and jogging.

All the trials were captured at 120 Hz for the Vicon system and at 60 Hz for the Xsens system and further process was done in MATLAB R2021a. We performed direct comparison over the whole time series of movement cycles and the Bland-Altman (BA) analysis [2] only at the clinically significant points. The clinically significant points for walking were defined at the moment of heel strike, mid stance, and initial swing of gait cycle [3] while the selected points for jogging were at the initial contact, loading response, and early float [4].

RESULT

Overall joint angular displacements of Xsens followed closely the trajectory of Vicon data. The root-mean-square difference (RMSD) between Xsens and Vicon data ranged from 2 to 15 degrees in the sagittal plane both for walking and jogging. The range of the minimum and maximum RMSD was smaller in the knee and the ankle joints compared to the hip joint in the sagittal plane. However, the RMSD value varies from 1 to 25 degrees and 2 to 40 degrees in the coronal and the transverse plane, respectively that is beyond the clinically significant limit (< 5 degrees). Figure 1 showed the result of the BA analysis of three clinically significant points in gait.

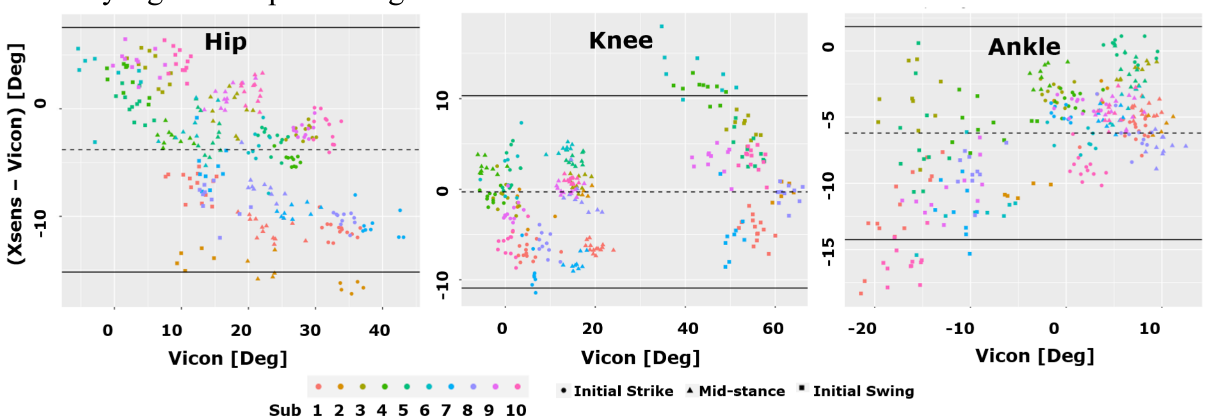

Figure 1: The Bland-Altman analysis resulted that difference between Xsens and Vicon in y-axis against Vicon in x-axis in the sagittal plane during gait. (Black-dot lines denote the mean of difference; solid lines notes ± 2SD.)

DISCUSSION

Under normal circumstances we would expect the errors identified in a Bland-Altman assessment to be random. However, on examination of our BA assessments we found in post processing a specific pattern of errors (differences) in Xsens while the distribution of errors in BA plots was supposed to be randomly scattered. We assumed that this pattern was highly correlated with previous measurement errors in each trial of Xsens. We found groupings of data about the Y-axis amongst individual participants, suggesting that the error detected between the systems was highly subject dependent. In addition, BA plots indicated that the difference was smaller when the measurement was a large angular movement although the larger difference was observed in a smaller angular movement.

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COMPARISON OF THEIA MARKERLESS VS. CONVENTIONAL GAIT MODEL

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INTRODUCTION: Marker-based motion analysis is the current standard for clinical gait analysis. Recently, commercial markerless motion capture systems such as Theia3D have become available and have the potential to provide more convenient technology for performing motion analysis. Studies have examined repeatability and concurrent validity of Theia3D in healthy adults [1, 2], but have not examined children or clinical patients with gait deviations. This study investigates concurrent validity of markerless gait analysis using Theia3D compared with the traditional Plug-in-Gait (PiG) model in pediatric patients undergoing clinical gait analysis.

CLINICAL SIGNIFICANCE: Markerless motion capture shows promise as an alternative to traditional marker-based gait analysis for children with gait disorders.

METHODS: Fourteen pediatric patients (7 male, age 5-16 years) with a range of diagnoses (Table 1) underwent clinical gait analysis with data being captured concurrently by a traditional marker-based motion capture system (PiG, Vicon, Oxford, UK) and a markerless system (Theia3D, Theia Markerless Inc., Kingston, ON, Canada). Multiple left strides from one or more walking trials were averaged for each subject, and the difference in kinematics (Theia - Vicon) was calculated over the gait cycle and evaluated using root mean square difference (RMSD), mean difference, and RMSD after subtracting the mean value across the gait cycle ($RMSD_{offset}$).

Table 1: Patient diagnoses and walking conditions

Diagnosis	N	Walking condition
Abnormal gait	4	4 barefoot (unassisted)
Cerebral palsy	4	1 barefoot (unassisted) 1 shod (unassisted) 1 braced (unassisted) 1 barefoot w/ crutches
Clubfoot	2	2 barefoot (unassisted)
Charcot-Marie-Tooth	1	1 barefoot (unassisted)
Spina bifida	2	1 barefoot (unassisted) 1 barefoot handheld & braced unassisted
Spinal cord AVM	1	1 barefoot (unassisted)

RESULTS: Kinematics showed similar patterns between the marker-based and markerless systems (Figure 1). RMSD was <5° except for pelvic tilt, hip flexion, ankle inversion, knee rotation, and ankle rotation (Table 1). These measures mainly differed due to a shift in the curves. After adjusting for this static offset, $RMSD_{offset}$ was ≤5.2°. The markerless system placed the pelvis in less anterior pelvic tilt than PiG, resulting in less hip flexion, likely due to a difference in definition of the pelvis neutral position, which is anteriorly tilted in the PiG model. There was also a large offset in knee and ankle rotation between the two systems, but little difference in foot progression (RMSD 1.6°, difference -1.2°) (Table 2, Figure 1). In some cases, knee varus showed a larger "bump" in swing with the markerless system, suggesting possible cross-talk between sagittal and frontal plane kinematics.

Table 2: Comparison between Theia and Vicon kinematics (all values in degrees)

	RMSD			RMSD after offset			Mean difference (Theia - Vicon)		
	X	Y	Z	X	Y	Z	X	Y	Z
Trunk	2.9	3.3	4.6	5.0	1.5	3.4	2.9	-3.3	4.5
Pelvis	11.9	1.5	3.6	0.5	1.0	1.7	-11.9	-1.1	3.1
Hip	13.1	2.3	1.9	1.3	2.3	1.9	-13.0	0.4	-0.3
Knee	1.7	2.4	10.9	1.7	2.4	2.9	-0.3	-0.4	-10.5
Ankle	1.5	7.0	17.0	0.9	5.2	1.7	-1.2	4.8	-16.9

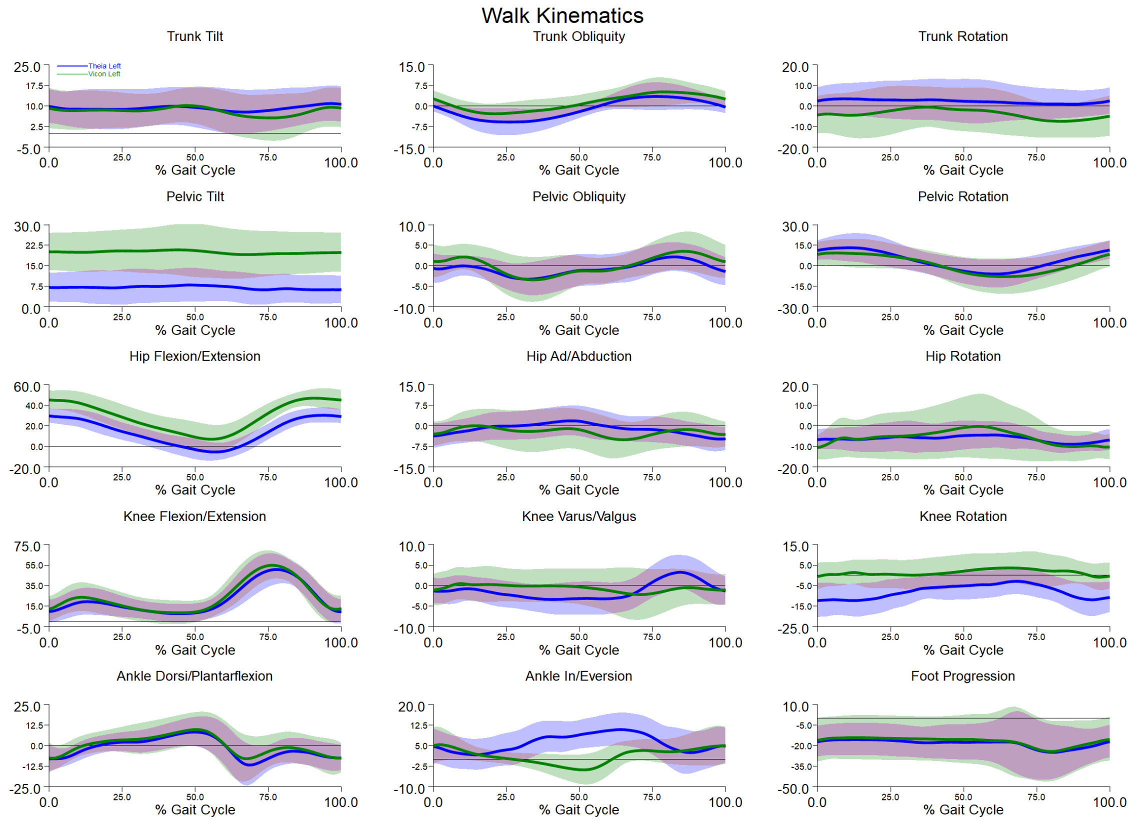

Figure 1: Comparison of average curves for Theia markerless (blue) and Vicon (green)

DISCUSSION: The markerless system generally produced similar kinematics to the traditional marker-based system. While differences in model definition need to be considered, these results provide preliminary evidence of concurrent validity of the markerless system for pediatric patients with abnormal gait. Larger studies are needed to confirm these results, especially for patients using braces or assistive devices. It should also be noted that it is unknown which system is more accurate due to lack of a gold standard such as imaging.

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THE HAZARDS OF BRIGHT LINE STATISTICS IN A WORLD OF UNCERTAIN DATA: A NEW TWIST ON HOLM'S METHOD

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INTRODUCTION

Functional data analysis of different sorts has recently come into vogue in biomechanical data processing. These methods address the problem of comparing groups of curves and determining where along the curves significant differences lie whilst also ensuring that excessive type I error is guarded against. Perhaps the most frequently cited of late has been Statistical Parametric Mapping using Random Field Theory (SPM-RFT), although there is active research into other related methods [1, 2]. Of late, however, the error control of SPM-RFT has been brought into question with some questioning the family-wise error rate [1], and others finding discrepancies between SPM-RFT results and those found using more conventional methods such as Holm-Bonferonni (HB) [2].

Juxtaposed against this is the issue of uncertainty in data which can include stride to stride, trial to trial, and day to day variability. Often, in the interest of keeping statistics simple, data are averaged within a subject to create a "representative" sample, thus not accounting for within subject uncertainty. Added to this is the fact that there are often relatively small subject pools. This can mean that statistical inferences are suggestive, but not bullet-proof until replicated by multiple labs. Bootstrapping is one method used to estimate population parameters in such cases by resampling the data with replacement a large number of times. This non-parametric method can give better results especially when data are not perfectly normal in their distribution. In particular, since the data are known to be uncertain on several levels, bootstrapped t-curves can be used to enhance the HB technique. Since this technique involves comparisons of points within curves to make statistical inferences, if the points within the curves are themselves uncertain, the HB results are likely to be spurious.

CLINICAL SIGNIFICANCE

This work demonstrates how different commonly used methods can yield dramatically different results when interpreted in the bright-line framework of significant or not, and offers guidance for interpreting results.

METHODS

Publically available medial gastrocnemius forces from Besier et al [3] were used for this illustration (Fig 1a). The data were comprised from two groups of subjects, 16 in a control group, and 26 in a pain group, and each subject was represented once with data given to a precision of tens of micro-Newtons on a hundred Newton scale (8 significant figures) and dilated to 100 points. Welch's method was used to obtain a representative t-value curve versus time. This curve was analyzed for significance using the HB method and SPM-RFT [4]. The data were then bootstrap sampled with replacement 1000 times to generate a family of t-curves for further analysis (Fig 1b).

A modified HB (HBM, or Holms-Bootstrap) method was then applied where the bootstrapped data were sorted from highest to lowest t-value, and then the bootstrap samples were compared to determine if there were significant differences between them. This resulted in another critical t-value and p-value for comparison to the prior methods.

DEMONSTRATION

In cases where the degrees of freedom are the same, approximate HB sorting can be done with t-values instead of p-values. If points along the t-curves were not significantly different, then they are equivalent according to the HB, and the effective number of tests is reduced. Table 1 presents results including percent of cycle that is significantly different between groups.

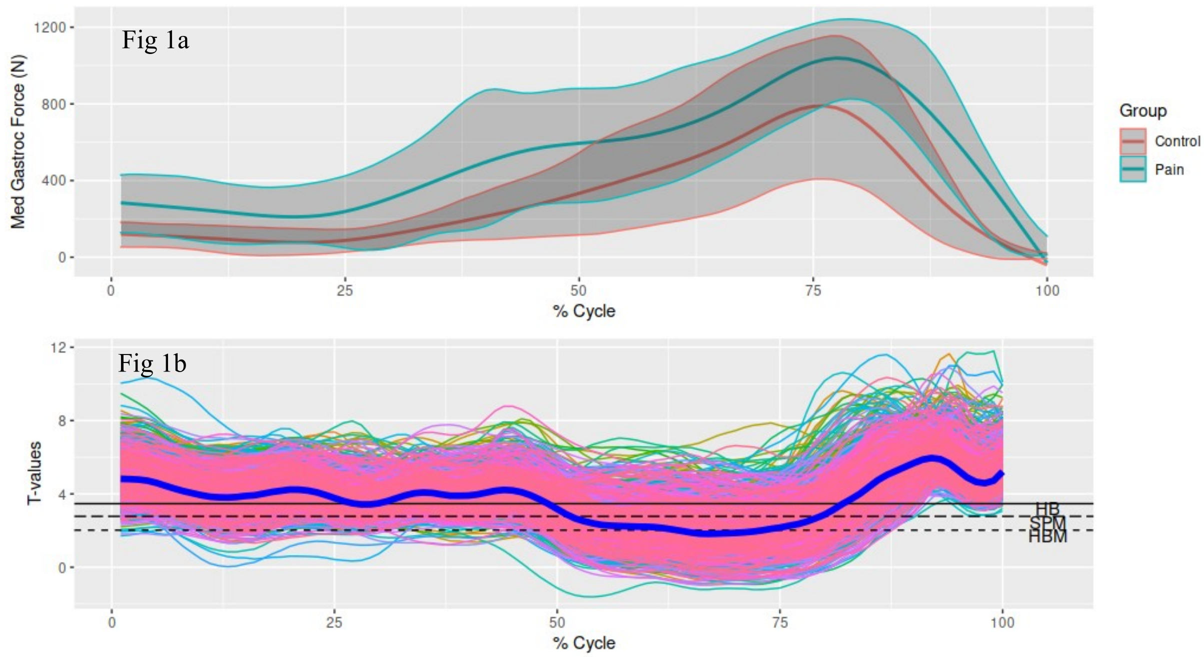

Figure 1a) Plot of mean data by group with grey standard deviation band. **1b)** Family of bootstrapped t-value curves corresponding to HBM with several bright line significance levels overlaid for comparison.

Table 1: Statistical summary

Method	t_{crit}	p_{crit}	**%cycle sig**	**Neff**
Bonferroni	3.78	0.0005	56%	100
Holm-Bonferroni (HB)	3.42	0.0014	65%	35
SPM-RFT	2.77	0.0083	70%	6
Holm-Bootstrap (HBM)	2.33	0.025	78%	2

Results indicate that the strict HB method is more conservative than the bootstrap which is in turn slightly more conservative than SPM-RFT. Given the uncertainty in the data the HBM method is the least conservative and indicates that there are two effective tests.

SUMMARY

The range of outcomes here demonstrate that different methods may yield widely different results, even between now venerable tests like HB and SPM-RFT. It is suggested that authors show multiple tests for significance to give the reader full context, or better yet stick with confidence intervals since bright-line significance testing under such circumstances is somewhat like measuring with a yard-stick and then cutting with a laser.

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DISCLOSURE STATEMENT

Timothy Niiler has no conflicts of interest to disclose.

CHANGES IN WHOLE-BODY AND SEGMENTAL ANGULAR MOMENTUM IN PERSONS WITH PARKINSON DISEASE ON AN IRREGULAR SURFACE

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INTRODUCTION

Whole-body angular momentum in relation to the body's center of mass (COM) is highly regulated in overground walking and is invariant to walking speed in healthy adults [1]. These conclusions are not generalizable to other conditions beyond level, overground walking such as negotiating 90-degree turns [2] as well as stair ascent and descent [3]. It is unknown whether conclusions regarding regulation of whole-body angular momentum in healthy individuals extend to populations that are neurologically impaired, such as those with Parkinson disease (PD). A recent investigation found that in stair descent, persons with PD regulated whole-body angular momentum differently than a healthy control population [4]. This finding, in addition to the growing body of literature showing that Persons with PD alter their kinematics and kinetics in the presence of irregular terrain [5] prompted us to investigate whether persons with PD regulate their whole-body angular momentum differently than healthy, age-matched control (HC) participants on a regular and an irregular surface.

CLINICAL SIGNIFICANCE

Impaired regulation of whole-body angular momentum is significant because rotational momentum that is not cancelled can lead to the COM exceeding the base of support. The COM exceeding the lateral border of the base of support can potentially lead to a fall [6]. In the event of a slip, poor regulation of angular momentum also increases slip severity [7]. This study identifies anatomical planes of movement and segments that should be targeted through therapy and/or external manipulation to aid regulation of angular momentum to reduce risk of falls.

METHODS

Nine persons with PD under dopaminergic medication (5 Males 4, Females; 67.7 ± 7.1 years; 1.66 ± 0.16 m; 81.0±20.6 kg; 36.1 ± 11.8 Unified Parkinson's Disease Rating Scale score; 2.39 ± 0.31 Hoehn & Yahr score) and nine healthy, age-matched controls (6 Males, 3 Females; 67.7 ± 8.0 years; 1.69 ± 0.05 m; 74.5 ± 5.6 kg) were recruited for this study. Participants ambulated on a custom 7.3 m long by 0.76 m wide walkway that had two possible surfaces: 1) <u>Regular</u>: oriented-strand board (OSB) and 2) <u>Irregular</u>: OSB covered in faux rock panel. A representative trial was selected from each participant on each surface to analyze. Outcomes were the whole-body angular momentum (WBAM) about the COM as well as the contributions of the head/trunk/pelvis (HTP), arms and legs segments in the sagittal and frontal planes. These metrics were determined from three-dimensional motion capture data and using Visual3D (C-Motion Inc, Germantown, MD). Positive sagittal angular momentum (X) was defined as rotating backward and positive frontal angular momentum (Y) was defined as leaning right. Data were normalized to participant height, mass, and average walking velocity. Surfaces and health groups were compared using a non-parametric statistical parametric mapping 2-way ANOVA with an alpha of 0.05.

RESULTS

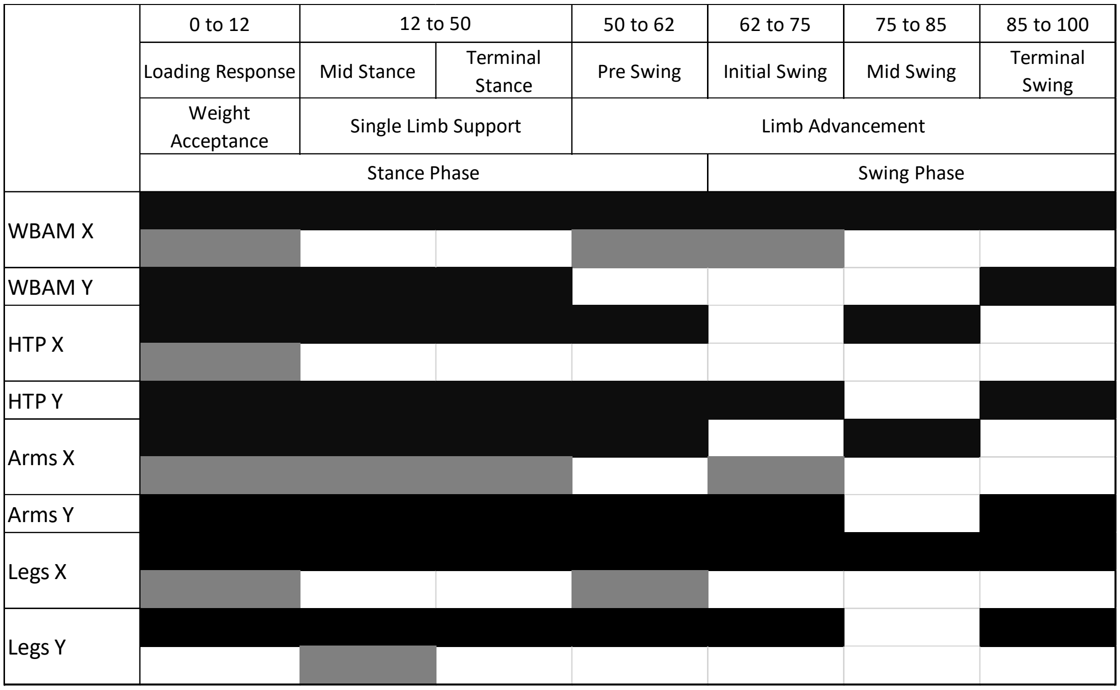

Figure 1: Results of 2-Way ANOVA. Solid black indicates significance between health groups. Solid grey indicates significance between surface type.

DISCUSSION

Our results show that the PD group always exhibited increased angular momentum compared to the HC group. There were notable differences in the sagittal plane where the PD group increased angular momentum during stance phase and decreased during swing phase. In the frontal plane, the PD group exhibited greater angular momentum during loading response, indicating they allowed their center of pressure to get closer to the edge of their base of support as compared to the HC group potentially making them more susceptible to lateral falls. The between surface results were inconsistent. Generally, both groups increased their sagittal and frontal angular momentum during loading response on the irregular surface. However, this was not always observed following the second step. On the irregular surface, both groups primarily regulated their sagittal angular momentum with their arms, indicating the importance of arm swing regulation, which is often impaired in persons with PD.

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DISCLOSURE STATEMENT

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EFFECTS OF GLOBAL SAGITTAL ALIGNMENT ON POSTURAL CONTROL SYNERGY IN RELAXED QUIET STANDING

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INTRODUCTION

Good postural alignment has primarily been described by empirical norming and/or negation of identifiable misalignment patterns. Such approaches are informative to a point, but likely fail to account for all clinically relevant variability in alignment, thus limiting the clinical utility of this modifiable behavioral factor. By contrast, affirmative and theoretically grounded accounts of good postural alignment may advance effective clinical application of posture-based interventions through increasing the precision of targeted alignment patterns and strategies.

Previous work has demonstrated that coordinative, posture-stabilizing joint angle synergies (S_N)—quantified using analytical frameworks based in referent control theory—tend to weaken as posture becomes more *mis*aligned.[1] These findings may suggest that some alignment patterns can be considered "better" than others on the basis of affirmative criteria and at the level of the individual. Even so, the research in question may not be generalizable to relaxed, quiet standing behavior for two reasons, both relating to the researchers' approach of using volitional leaning to induce misalignment. First, the scale of misalignment likely exceeded what would otherwise be observed in relaxed standing. Second, intention to lean would be achieved through a shift in the referent control parameters underlying the applied analytical methods, potentially confounding their results.

The purpose of this study was to extend previous findings on the relationship between S_N and misalignment to experimental tasks of greater ecological validity. We hypothesized that deviation from individually identified postural alignment configurations would associate with decreasing S_N.

CLINICAL SIGNIFICANCE

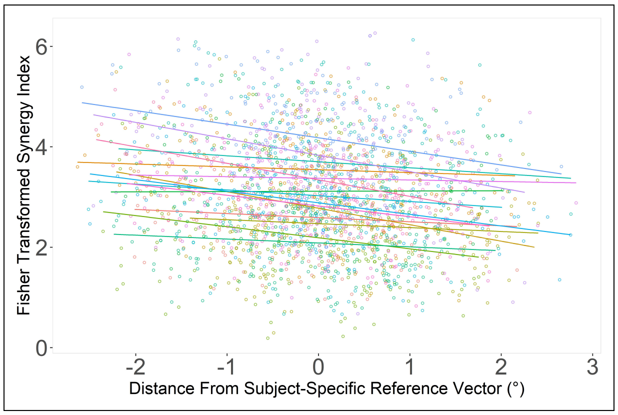
The lack of consistent definitions of good postural alignment limits the efficacy of posture assessment and intervention. The present study suggests that motor control theory could lead us to more precise definitions of good postural alignment. In the future, such affirmative postural targets may establish a basis for shifting the focus of postural intervention from avoiding "bad" alignment to promoting "good" alignment.

METHODS

Fifteen adults (24.53 ± 3.54 years, 168.97 ± 9.70 cm, 74.75 ± 13.18 kg) completed 4-min. quiet standing trials in five different foot placement conditions: 1) control, 2) 30° toes-in, 3) 30° toes-out, 4) 12° toes-up, 5) 12° toes-down. Sagittal plane segment (foot, tibia, femur, lumbar spine, thoracic spine, neck/head) and joint (ankle, knee, hip, thoracolumbar junction, neck) angles were calculated from surface marker trajectories acquired at 100 Hz (VICON, Oxford, UK). Trials were then segmented into nonoverlapping 15-sec. windows. For each window, we applied the uncontrolled manifold framework to calculate a normalized Fishertransformed synergy index (i.e. S_N), which quantified the tendency of the observed joint angle variability to stabilize anteroposterior displacement of the center of mass. For each subject, a reference configuration of segment angles (excluding the foot segment) was determined as the trial-wise average of the synergy-maximizing stance condition. Finally, a misalignment (M) value was calculated as the mean Euclidean distance from the reference configuration for each 15-sec. window of each trial for a given subject. The effects of M on S_N were assessed using the "Random-Effects Within-Between" (REWB) [2] model at an *a priori* α of 0.05.

RESULTS

Fig. 1



S_N (y-axis) plotted against person-mean-SD standardized Euclidean distance from referent configuration (x-axis). Greater x-axis values indicate greater Euclidean distance from a participant's reference configuration. Each point represents a single 15-sec. window within a trial. Each line represents a linear model fit across all data for a given participant.

Table 1. REWB Model Summary

Intercept			M_W			M_B		
β	*t*	*p*	*β*	*t*	*p*	*β*	*t*	*p*
2.58	4.87	<0.001	-0.16	-5.73	<0.001	0.12	0.96	0.36

Nakagawa conditional and marginal R^2 were 0.31 and 0.04, respectively. A significant within-subjects effect of misalignment (M_W) on S_N was observed (Table 1). On average, S_N decreased as individuals deviated further from their respective reference configurations (Fig. 1).

DISCUSSION

We studied the effects of misalignment on an affirmative indicator of postural control quality. Our findings support the notions that 1) it may be possible to identify affirmative postural targets, and 2) meaningful variability in alignment may be measurable on an individual basis and on scales relevant to relaxed, quiet standing.

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DISCLOSURE STATEMENT

The authors have no conflicts of interest to disclose.

IMPACT OF INSTRUCTION TYPE ON BALANCE STRATEGY AND PERFORMANCE OF INDIVIDUALS ON A PASSIVELY UNSTABLE SURFACE

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INTRODUCTION

Functional and occupational tasks often involve interacting with a passively unstable surface (PUS). For this area of investigation, a PUS, is defined as a non-fixed surface that moves in response to movements of an individual standing on it [1]. Such surfaces can include small boats, loose gravel, scaffolding, etc. Maintaining balance on PUS's can be especially important for older adults, athletes, and certain worker types (e.g., construction workers), since falls by such persons can be particularly problematic.

Previous investigation of balance on PUS's has identified effects, on balance performance and mechanisms, of task related factors such as impairment of visual acuity, presence of a concurrent task, and degree of surface instability [2-4]. However, it is conceivable that variation of intended balance task objectives may also impact balance performance and mechanisms. Tasks performed on PUS's across varying domains can be associated with different balance objectives. Precisely placing objects, when standing on thick foam, may necessitate that upper body segments remain very still. Conversely, concern regarding potentially capsizing a small boat or simply disturbing other passengers, may dictate keeping the standing surface very still to be the primary objective.

In PUS balance research, subjects are often instructed to keep the surface still. However, keeping their bodies still may be a more relevant instruction for many applications. Impacts of these different instructions on balance performance and strategy requires attention. Individuals might possibly interpret both instructions similarly, or they may alter balance strategy in response to instruction type. The purpose of this research was to determine whether instruction type affects balance performance and strategy of subjects on PUS's, and whether instruction type mitigates impacts of varying visual acuity, concurrent cognitive demand, and external PUS support.

CLINICAL SIGNIFICANCE

Understanding, of how instructions regarding PUS balance task objectives can affect balance strategy, could be utilized in balance assessment and balance training applications. Directing a person's focus towards keeping the body still, rather than keeping a PUS still, may improve overall stability by reducing body segment movements, while not increasing PUS motion.

METHODS

Ten healthy young subjects (4 m, 6 f; 21-23 years) participated, after providing informed consent. For each trial, subjects stood on a wobble board for 45 seconds, while instructed to either keep the wobble board as still as possible (Board Still) or their body as still as possible (Body Still). Five subjects received Board Still first and performed 2 trials with each permutation of intact vision/visual blur (IV/VB), no concurrent cognitive task/verbal fluency task (NC/VF), and no foam under wobble board/foam support under wobble board (NF/FS). These subjects then received Body Still and repeated all trial permutations. The remaining subjects performed trial permutations with instruction order reversed. Trajectories of reflective markers, placed on subject body segments and on the wobble board surface, were obtained with an optoelectronic motion tracking system (Qualisys) at 100 Hz. Body segment metrics, computed from marker coordinates, were mean angular velocities (AVs) for the trunk, pelvis, upper arms and lower arms. Metrics computed for the wobble board were mean AV and tilt angle standard deviation (TiltSD).

RESULTS

Mean upper arm, lower arm, trunk and pelvis AVs were significantly lower with Body Still instruction (Fig 1). Instruction type did not significantly impact board mean AV or TiltSD (Fig 2). All AVs and TiltSD were significantly higher with VB and with NF (Figs 1 and 2). Mean upper arm, lower arm, trunk and board AVs were significantly higher with a VF task (Figs 1 and 2). A VF task did not significantly impact mean pelvis AV or board TiltSD (Figs 1 and 2). Instruction type influence on external support impact was significant for trunk, upper arm, and lower arm mean AVs, such that increases with NF were lower with Body Still instruction. Instruction type influence on cognitive demand impact was significant for upper arm and lower arm mean AVs, such that increases with VB were present with Body Still instruction.

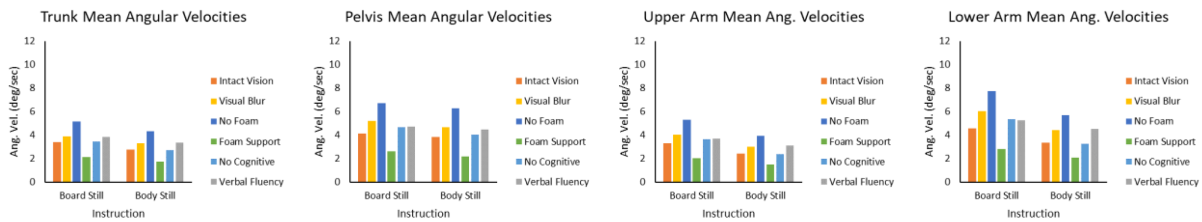

Fig 1. Body segment mean angular velocities.

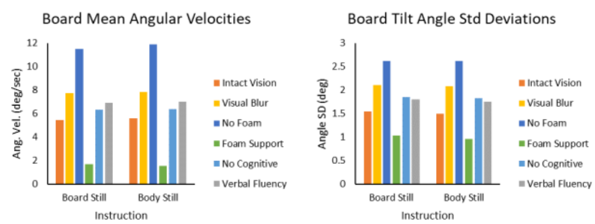

Fig 2. Wobble board movement metrics.

DISCUSSION

During a PUS balance task, decreased body segment movement with Body Still instruction is not particularly surprising. It may simply indicate persons performing tasks as asked. Of more interest may be lack of increased PUS movement with Body Still instruction, such that improved (i.e., reduced) body segment movement did not incur a cost of greater PUS motion. Significant interaction influence, of instruction type on external support effects, suggests that Body Still instruction may be particularly beneficial with more challenging PUS's. A potential drawback of Body Still instruction is demonstrated by the significant interaction influence of instruction type on cognitive demand effects. Purposeful execution of Body Still instruction may invoke greater cognitive effort, which may then incur greater negative impact from an additional cognitive task.

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DISCLOSURE STATEMENT

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Is Overall Gait Performance Or Rate Of Force Development Associated With Fall History in Adults With Cerebral Palsy?

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Introduction

Falls are the second leading cause of unintentional injury deaths worldwide, in which it is estimated that a total of 37.3 million falls lead to severe injuries that require medical attention and roughly 684,000 mortalities each year [1]. This can be financial costly, especially in frail populations (adults older than 65 or individuals with disabilities), in which roughly $50 billion is spent on medical costs related to non-fatal fall injuries and $754 million is spent related to fatal falls each year [2].

To reduce the costs of treatment for falls, fall risk tools, questionnaires, functional ability assessments, and prevention programs have been developed to reduce the number of falls [3]. Within the geriatric population, specific biomarkers to identify people at risk for falling have been studied, that includes balance impairment, decreased muscle strength, visual impairment, gait impairment/difficulties, depression, and functional motor limitations [4].

While falls and fall risk factors have been extensively studied in the geriatric and stroke populations, there is limited knowledge on the specific risk factors for adults with cerebral palsy (CP). Furthermore, it is unknown whether biomarkers commonly used to assess physical performance in individuals with CP (rate of force development-RFD, average gait speed, Gait Deviation Index-GDI) are associated with falls in this population. The goal of this study was to determine if these measures are associated with fall frequency in a cohort of adults with CP.

Clinical Significance

Understanding the association between the incidence of falls and common physical performance measures related to rate of force production, gait performance and gait pattern or quality in a group of adults with CP can help us identify individuals at greater risk for falls in the future. This can also help us understand the importance of maintaining walking ability into adulthood in this population.

Methods

The current study represents a secondary analysis of data collected as part of the Cerebral Palsy Adult Transition Study (CPAT), in which the methodology has been previously published. A total of 72 ambulatory adults with CP were identified from a patient registry and were enrolled into this study. Demographical and medical history data were collected using questionnaires. Physical activity levels were assessed using the PROMIS-57 physical function domain. A train physical therapist with over 15 years of experience with individuals with CP assessed spasticity at the hamstrings, quadriceps, peroneals, posterior tibialis, toe flexors and gastrocnemius using the modified Ashworth scale. Peak RFD for both quadriceps were collected using a Humac Norm isokinetic testing machine set to evaluate isometric strength and rate of force development at 60 degrees of knee flexion bilaterally. An instrumented gait analysis was used to assess participant's average gait speed and GDI. Fall frequency (# of times a patient fell within a year) and fall severity (whether a fall resulted the participant to visit the emergency room, hospital, or doctor) were assessed utilizing the HELPS TBI screening tool. A multivariate negative binomial and logistical regression were used to assess associations between fall frequency and fall severity, respectively, with peak RFD, GDI, and gait speed.

Results

After adjusting for spasticity, sex, ethnicity, and GMFCS level, it was found that GDI and gait speed were statistically associated with fall frequency, in which for every one-unit increase in GDI and 1 m/s increase in gait speed, the number of falls per year would decrease by 0.0464 (95% CI: decrease of 0.0889 to 0.0039) and by 0.0705 (95% CI: decrease of 0.1222 to 0.0188) respectively. Peak RFD was not statistically significant. After adjusting for spasticity, it was found that GDI was statistically associated with fall severity (p-value=0.047), in which on average, the odds of having a severe fall decreases by 6% (95% CI: decrease of 11.6% to 0.1%) for every 1 unit increase in GDI. Walking speed adjusted for height and average peak RFD was not statistically associated with fall severity.

Discussion and Conclusion

This study provides evidence that gait performance is associated with fall history, but not peak RFD. Furthermore, it appears that gait pattern, as measured by GDI, is the best indicator for fall history, as it is associated with both fall frequency and fall severity. When interpreting the fall frequency results, it was found that to decrease the number of falls per year by 1, GDI would need to increase by roughly 20 units or the gait speed increase by 14 statures/min. Since this level of change represents large increases in gait performance, it might not be reasonable to improve the gait ability, just to see a small improvement in fall frequency. When interpreting the fall severity results, it was found that when an adult with CP has a clinically relevant improvement in GDI (increase of GDI by 5 units), the odds of having a severe fall decreases by 30% (95% CI: decrease of 58% to 0.5%). Since adults with CP are a heighten risk for falling, in which our cohort experienced a mean of 57.35 (95% CI: 15.05 to 99.65) falls within one year, decreasing the risk of having a severe fall is extremely important, as adults with CP have an increased number of falls. As a result, improving overall gait performance, specifically the pattern of movement represented by GDI, can significantly decrease the risk of having a severe fall within the adult CP population. Therefore, clinicians should try to maintain gait performance in ambulatory adults with CP, as this would not only improve the overall health of these individuals, but it would help them maintain their independence and reduce the costs to keep these individual healthy.

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Comparing Markerless And Marker-Based Motion Capture Systems For Clinical Motion Analysis Of Children And Youth With Atypical Lower Limb Mobility

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PURPOSE: The purpose of this tutorial is for attendees to learn about pros and cons of markerless motion capture approaches (specifically, Theia3D), with particular focus on the discrepancies between markerless and marker-based approaches.

INTENDED AUDIENCE: This tutorial is intended for anyone utilizing, or considering utilizing motion capture for clinical motion analysis in the pediatric population; this may include engineers, kinesiologists, physiotherapists, occupational therapists, prostheticians and orthotists, physiatrists and orthopedic surgeons.

PREREQUISITE KNOWLEDGE: This tutorial will utilize typical kinematic representations of gait and clinical motion in graphical format. Familiarity with these graphical representations of motion as well as the underlying theory that covers segment coordinate systems and calculation of joint angles, is recommended.

COURSE OUTLINE:

Introduction: Gait and clinical motion analysis has traditionally relied on marker-based motion capture systems to collect data, including kinematics. However, limitations associated with the placement of the markers – including accurate placement of anatomical markers and repeatability between operators – have been well recognized by researchers in this field for some time. Furthermore, the amount of hardware required for capturing marker-based data, the amount of time required to affix and remove markers, and the effect that marker placement has on the ability to capture 'unencumbered' motion data, are all limitations that are currently accepted as necessary [1]. Markerless motion analysis systems are becoming more and more popular, and have already been concurrently compared, and presented as a potential reasonable alternative, to marker-based systems for clinical motion analysis in specific patient populations [2]. Gait and Clinical Motion laboratories are often focused towards pediatric biomechanics, thus, there is an opportunity for further evaluation of novel markerless motion analysis systems in pediatric populations involving: those with atypical gait, those who use walking aids, and those recovering from interventions and/or injury.

While previous work by Kanko et al. have quantified comparisons between marker-based and markerless systems [2], the inherent functioning of the pose estimation algorithms, utilizing machine learning algorithms, of the Theia3D markerless system has the potential to perform differently in cases where the subject in the video exhibits atypical anatomy or physiological motion (e.g. gait, sporting maneuvers, etc.). We intend to survey how well marker-based and markerless system analyses perform across a variety of patients we see in our lab. Our aim will be to identify characteristics – anatomical, mobility-aid and motion-based – that affect the performance of the markerless system.

Clinical Significance: Changing the way motion analysis data are collected, without markers, could have a significant role to play in moving the field of motion capture forward. There is potential for a systemic change in clinical resource use, patient/family time required and overall quality of experience.

Methods: At The Motion Lab, Sunny Hill Health Centre at BC Children's Hospital (TML), we have been capturing and analyzing gait and clinical motion trials with both a Qualisys marker-based system (Qualisys, Göteborg, Sweden) and a Theia3D markerless system (Theia Markerless Inc., Kingston, Canada). We have established a routine process in which to collect, analyze and compare (using Visual3D) gait and clinical motion data of our patients.

Demonstration: In this session, we intend to demonstrate our process for collecting and analyzing data, so attendees can get a sense of what processes are involved with using two systems simultaneously, but with specific focus on the utilization of a Theis3D markerless system. We also will be presenting a variety of case data (example: Figure 1) to facilitate discussion regarding when the two system analyses agree, and when they do not.

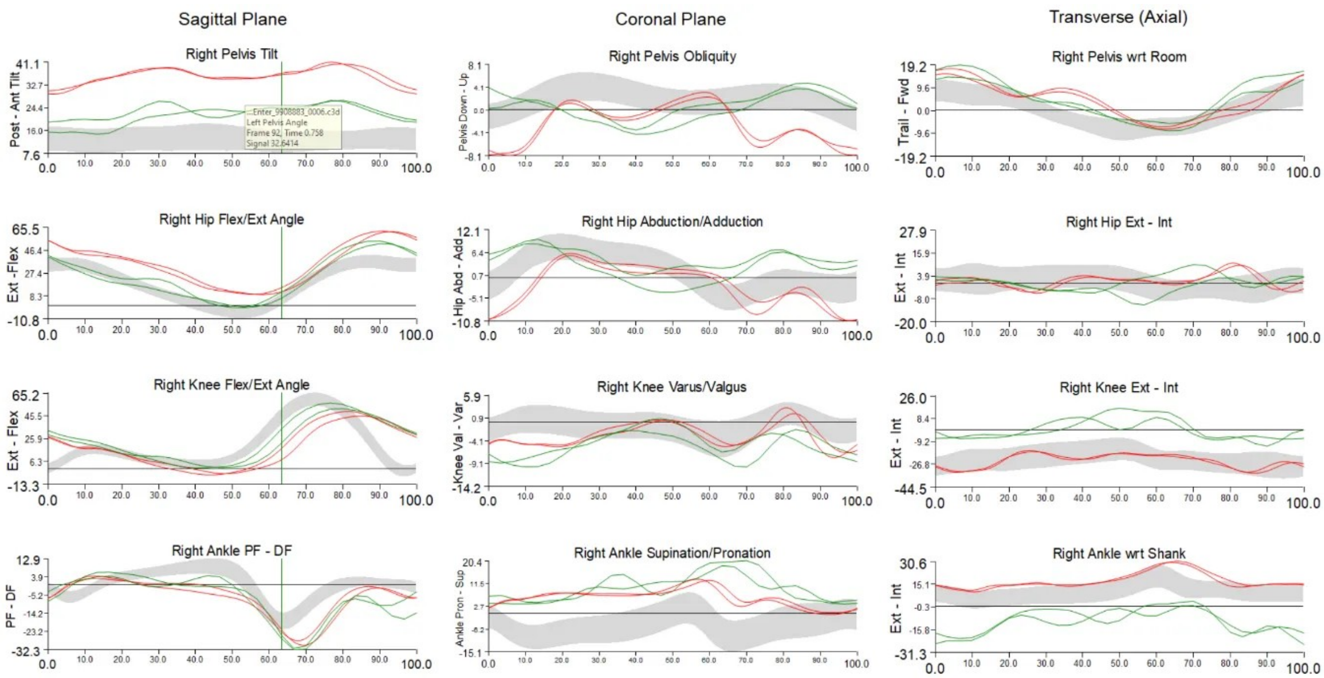

Figure 1 – Kinematic comparison data (n=1) between marker-based (RED) and markerless (GREEN) analysis. Sagittal, coronal and transverse kinematics are in the left, center and right columns, respectively.

Summary: Markerless motion capture systems are garnering a lot of attention for biomechanical analysis purposes. In this session, we intend to demonstrate our simultaneous use of a marker-based and markerless system to collect gait and clinical motion data from pediatric patients. We will also discuss discrepancies in the analyses from each system, with reference to distinguishable features of the patients.

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COP MEASURES OF POSTURAL SWAY DURING TANDEM STANCE UNDER EYES OPEN- AND CLOSED-CONDITIONS: A PILOT STUDY COMPARING NON-CONTACT SPORT ATHLETES AND FOOTBALL PLAYERS

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INTRODUCTION

The prevalence of sports-related concussions ranges from 1.6-3.8 million per year, with 50-80% unreported incidences [1]. Sports-related concussions result in subacute/chronic deficits and functional limitations, including the control of balance, with increased risk for subsequent musculoskeletal injuries [1]. Research purports that many athletes return to play (RTP) prior to the resolution of full central nervous system (CNS) recovery, suggesting that current RTP protocols are insufficient [1]. The inverted pendulum model has been used to describe the control of the center of mass during bipedal stance [2]. The medio-lateral (M/L) and antero-posterior (A/P) excursions and velocities of the center of pressure (COP) are commonly used to quantify postural control in non-contact (NC) and contact sport (CON) athletes [3]. However, there is limited information on these COP metrics for tandem stance postural control in healthy and concussed athletes [4]. Also, there is limited information regarding the effect of leg dominance [5], and eyes-open (EO) and eyes-closed (EC) testing conditions [6]. Thus, the purpose of this study was to measure COP range and mean velocity in the M/L and A/P directions in collegiate NC sport and football athletes in tandem stance postures, under EO and EC conditions.

CLINICAL SIGNIFICANCE

Our findings are consistent with previous studies that have shown the importance of the visual system in postural control. Moreover, our results suggest that potential differences in COP metrics between types of athletes should be considered when creating control data sets to be used when testing concussed athletes. Within the limits of our project, we conclude that there appears to be a need to include COP metrics as part of RTP decisions for concussed athletes.

METHOD

Ethics approval was obtained from the Grand Valley State University Human Research Review Committee (IRB# 17-102-H). CON sport athletes including 19 Division II football players (18.5±0.5 yrs.; height 186.0±6.0 cm; mass 110.9±22.4 kg; BMI 31.8+5.9 kg/m^2) and 17 NC sport athletes (19.4±0.5 yrs.; height 192.4±6.2 cm; mass 76.3±9.5; BMI 22.9±2.0) participated. Ground reaction force data were collected (200 Hz) using AMTI NetForce software (Advanced Mechanical Technology, Inc., Watertown, MA), as participants stood with one foot in front of the other, with the back foot on the force plate, and the heel of the front foot placed on a mark approximately 1 cm in front of the force plate. Athletes were instructed to stand with equal weight on each limb, hips, and knees extended, with arms crossed in front of the chest, and to look straight ahead. Participants stood on the force plate with the foot aligned along the plate's x-axis, thus COP movement along the x-axis represented the A/P COP excursion. Five good trials (no significant loss of balance or opening of eyes) were collected for 10 seconds for each foot on the force plate under two conditions (EO and EC), for a total of 20 trials. A custom MATLAB program was used to extract raw GRF data (filtered with a 2nd-order lowpass Butterworth filter cutoff at 5 Hz) and determine the dependent variables: maximum M/L and A/P

COP range and M/L and A/P COP mean velocity. SAS JMP (SAS, Cary, NC) was used to determine descriptive measures of participant demographics and COP variables. A mixed-model ANOVA was used to analyze data with $p < 0.05$ to determine significance, with Bonferroni corrections. Primary response variables were analyzed with a three-factor design ANOVA with factors of sport, eye condition, and limb. Random factors of participant within sport and leg within-participant were included. Main effects and the two-way interaction between sport and eye conditions were also analyzed. A Tukey's post hoc HSD analysis was used as needed. The effect size was defined as the difference between the means of the two groups defined by the factor levels after correcting for the other factors in the model. Effect sizes for the COP range were reported using the units of the dependent variables, and defined as large (≥30 mm), medium (11-20 mm), and small (≤10 mm), based on the clinical judgment of the researchers.

RESULTS & DISCUSSION

Variation in dependent variables between trials was not significant and there was no difference in COP range and mean COP velocity between limbs. Likewise, there was no significant interaction between type of sport and eye condition. However, within groups, there was a greater mean COP range and velocity in both M/L and A/P directions in the EC condition ($p<0.0001$) (Table 1). Between groups, a significant interaction between type of sport and eye condition was found in the mean COP velocity M/L direction ($p=0.058$).

Table 1. COP Range in Anterior/Posterior and Medial/ Lateral Directions with Varying Conditions (Sport Type, Limb Position, and Eyes Open or Closed).

			Mean (±SD)	Median	CV (%)	Range (Min-Max)	P-Value	Effect Size
COP A/P Range	Sport	Football Athletes	42.94 (26.98)	34.56	62.82	174.49 (0.14-174.63)	0.2117	
		Non-Contact Athletes	50.36 (30.40)	42.11	60.37	152.65 (0.80-153.50)		
	Eyes	Eyes Closed	59.28 (32.83)	50.71	55.38	174.49 (0.14-174.63)	<0.0001*	25.36
		Eyes Open	33.87 (16.57)	31.33	48.92	125.46 (0.80-126.27)		
COP M/L Range	Sport	Football Athletes	32.04 (14.96)	31.35	46.70	75.68 (0.04-75.72)	0.1707	
		Non-Contact Athletes	36.31(15.05)	35.04	41.45	101.60 (0.55-102.23)		
	Eyes	Eyes Closed	42.65 (14.97)	42.29	35.10	102.19 (0.04-102.23)	<0.0001*	17.06
		Eyes Open	25.62 (9.51)	24.73	37.11	70.45 (0.55-71.00)		

* Indicates significant difference with $p<0.05$ with Bonferroni correction of 0.0125
Data displayed in Millimeters
Coefficient of Variation = CV (%)

Although there are discrepant findings in the literature, our finding of the effects of EC conditions on postural sway was consistent with previous reports [6,7], suggesting the importance of the visual system in postural control. As reported by others, limb dominance does not appear to be a factor in postural control [5].

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KINEMATIC COMPARISON BETWEEN SNATCH AND CLEAN TECHNIQUE IN WEIGHTLIFTING

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INTRODUCTION

Weightlifting consists of the snatch and the clean and jerk [1]. Both techniques start with the same movements until the second pull phase, then differentiated during turnover, catch, and recovery phases [2]. Most previous studies have separately investigated each technique in terms of kinematic, kinetic as well as mechanics of the barbell. These two techniques have distinct movement sequences but lack information on joint kinematics during the snatch and clean techniques. In this study, we compared the shoulder, elbow, hip, and knee joint movements of the weightlifters during the snatch and clean lifts using the inertial measurement unit (IMU) sensors.

CLINICAL SIGNIFICANCE

This study compared upper and lower extremity kinematics during the snatch and clean lifts using the wearable inertial sensors which may help to choose appropriate exercise and training strategies for two main techniques.

METHODS

In this study, seven female Mongolian weightlifters participated. Two test sessions were carried out to record upper and lower extremity movements. The first sessions were snatch followed by clean. Each participant performed three snatch and clean lifts at 70 % of the one-repetition maximum. The shoulder, elbow, hip, and knee joint data were recorded using the IMU sensors (Wearnotch, Notch Interface Inc), which were attached to the chest, waist, arm, forearm, thigh, and shank using the straps. The joint angles were calculated, normalized and averaged for two different technique using Matlab® [3]. The full phases of the snatch and clean were defined [1,2,4] (Figure 1). An independent sample t-test was used to compare the mean values of joint angles during the snatch and clean with a significant level of 0.05.

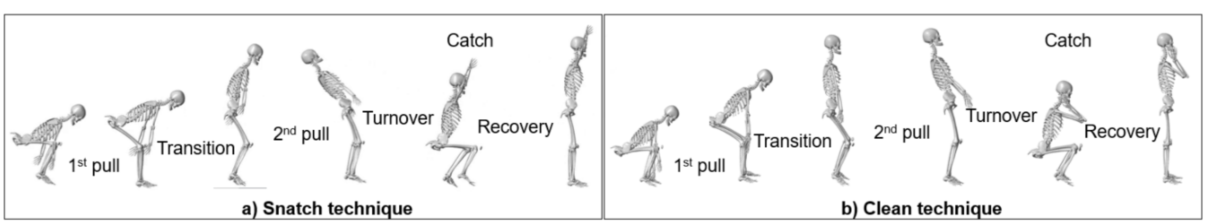

Figure 1. Full phases of the a) snatch and b) clean techniques

RESULTS

The maximum knee, hip, shoulder, and elbow angles were 125°/157°, 137°/171°, 165°/89° and 185°/172° in the snatch/clean, respectively (Figure 2). A statistical test showed a significant difference between the snatch and clean techniques in shoulder flexion ($p<0.05$) and elbow extension ($p<0.05$). Although the hip and knee joints were more extended during the clean when compared to the snatch, there was no significant difference was observed.

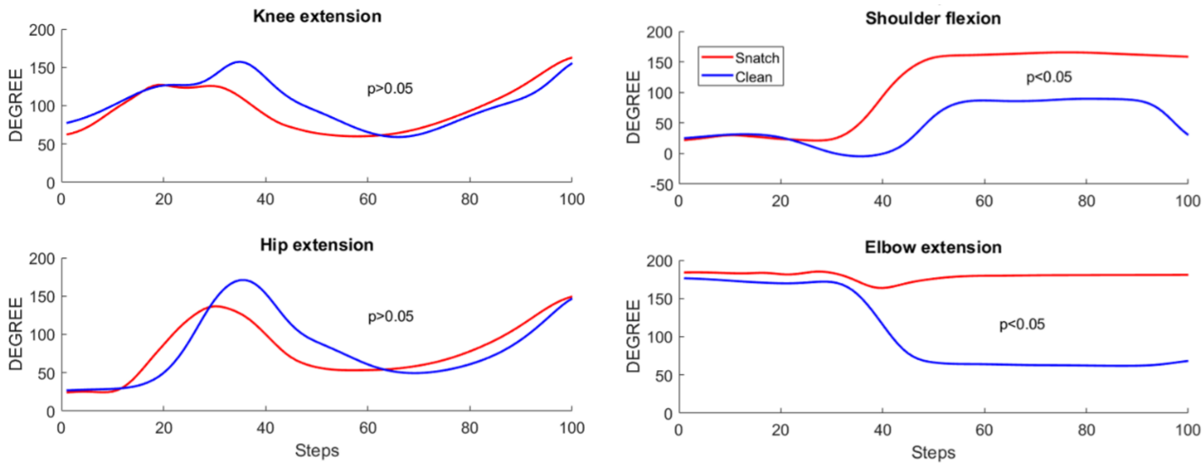

Figure 2. Joint angles during the snatch and clean techniques with p values

DISCUSSION

In this study, we compared the upper and lower extremity joint angles between the snatch and clean lifts. The shoulder and elbow movements were significantly different while the hip and knee movements were similar. It is clear that arm movements are different in two techniques; such as lifters moving the barbell from the floor to over-head in the snatch and to shoulder in the clean [1]. Similar hip and knee extension patterns were observed which is known to elicit the greatest amount of power output in both snatch and clean during the pulling phases [5]. Although this information is known to coaches and athletes, it is better to be aware of the detailed body segment movements with the advance of motion analysis technology. This study provides a comparative analysis of the joint kinematics between the snatch and clean technique, which may help to deeply understand similarities between the two techniques and enable to best prescribe the appropriate exercises.

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DISCLOSURE STATEMENT

The authors have no conflicts of interest to disclose.

GENDER COMPARISON OF GROUND REACTION FORCE AND CENTER OF MASS DURING OVERGROUND RUNNING IN TYPICAL ATHLETIC ADOLESCENTS

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INTRODUCTION

Kinetics and kinematics of recreational and elite runners has been extensively studied in the literature [1] with limited research examining the adolescent population. Adolescent runners have been evaluated using 3D motion analysis in one study comparing treadmill and overground running [2]. Compared to adults, adolescents had a greater center of mass (COM) excursion during treadmill running [3]. Studies have focused on factors associated with running injuries and injury rates to improve performance. Runners with stress fractures history were found to have high vertical ground reaction force (vGRF) loading rates [4] and a lower step rate has been associated with a decreased risk of bone stress injuries in college runners [5]. Further, adult female runners have shown higher vGRF loading rates than adult male runners, though this did not affect the incidence of running injuries [6]. Further, a higher vGRF loading rate was found in injured female runners compared to non-injured [7]. The purpose of this study was to evaluate vGRF and COM height during overground running in a healthy, adolescent athletic population.

CLINICAL SIGNIFICANCE

Understanding typical motion patterns in adolescent runners will help develop evidence-based assessment, training, and rehabilitation protocols specific to this population.

METHODS

Overground running data was collected as part of a larger IRB-approved study. Wearing personal shoes, athletes ran at a self-selected speed on a 32-meter path, with 9.75 meters available to gain speed and achieve their stride prior to crossing in-floor force plates. 3D motion data were collected using a 12-camera Vicon MX system (VICON, Los Angeles, CA) at 120 Hz, while simultaneous vGRF were collected using 2 in-floor force plates (AMTI, Watertown, MA) at 1080 Hz, following standard procedures [8]. A single running trial with a clean force plate strike of the preferred kicking limb was chosen as a representative trial for each subject to be used for analysis. Descriptive statistics were reported, and Mann-Whitney tests were used to determine differences when grouped by gender.

RESULTS

Data from 41 adolescent athletes (24 male, 17 female) with a mean age of 15.0 ± 1.8 years were included. Mean speed was 4.03 ± 0.4 meters/second. Median values for selected variables were analyzed (Table 1). Height and cadence were found to be statistically different between males and females. All other variables were not found to be statistically different between males and females.

DISCUSSION

Our study found a significantly higher cadence during overground running in adolescent females than males. Height was also found to be significantly lower in females. Given a shorter height, female runners may use a higher cadence to achieve the same speed as male runners. Additional analysis with normalization of cadence to leg lengths would further clarify this finding. A higher cadence in males and a lower cadence in females was found compared to published findings of college runners during treadmill running [5]. Our findings did not identify a difference in ground reaction force loading rate between adolescent males and females which is contrary to reported findings of higher loading rates in adult females [6]. As vGRF loading rate has been associated with greater incidence of stress fractures [4] and in previously injured female runners [7], further investigation of this relationship in the adolescent population is needed.

Table 1: Gender comparison of median values for selected variables

Variable	**Females (n=17)**	**Males (n=24)**	***p* value**
Age (years)	15.6	14.4	.49
Height (m)	1.63	1.73	.01*
Body mass (kg)	56.4	61.1	.33
Speed (m/s)	4.0	4.1	.34
Cadence (steps/minute)	175.6	167.4	.024*
Vertical GRF Loading Rate (N/s)	23.79	26.62	.37
Maximum GRF in stance (% Bodyweight)	4.6	4.5	.81
COM height at foot strike (% Height)	0.54	0.54	.65
COM height at peak knee flexion (% Height)	0.51	0.51	.46
COM height at foot off (% Height)	0.54	0.55	.56
COM displacement (% Height)	0.04	0.043	.29

*Significance set at $p<0.05$. vGRF= vertical ground reaction force, COM=center of mass

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DISCLOSURE STATEMENT

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DISTAL FOOT INITIAL CONTACT AND PEAK ANGULAR VELOCITY DURING RUNNING IN RUNNERS WITH PLANTAR HEEL PAIN

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INTRODUCTION

Running is one of the most popular recreational activities in the world, however, it is also associated with a wide range of injuries [1]. One of the most common injuries is plantar heel pain [2]. Despite the prevalence and the fact that plantar heel pain is believed to be caused by mechanical overload of the plantar fascia, few studies have investigated distal foot kinematics during running. Thus, we hypothesized that distal foot angular velocity would be increased at initial contact and during the stance phase of running in runners with plantar heel pain.

CLINICAL SIGNIFICANCE

Increased angular velocity within the distal foot during running may lead to, and/or be the result of, degeneration of the plantar fascia and the intrinsic foot muscle weakness believed to be associated with plantar heel pain [3].

METHODS

Twenty-six runners participated in the study. Plantar heel pain participants (PHP) (7 F/6 M, Age: 29 ± 8.3 y., BMI: 23.1 ± 2.0 kg/m^2) complained of clinical plantar heel pain symptoms for at least six weeks prior to the study. Uninjured participants (CON) (6 F/7 M, Age: 30.5 ± 5.9 y., BMI: 23.5 ± 3.1 kg/m^2) were age, sex, and BMI matched with the PHP group. A seven-segment foot model that defined six functional articulations (rearfoot, medial and lateral midfoot, medial, and lateral forefoot, 1st metatarsophalangeal) was used to quantify foot motion [4]. Following foot model marker application, participants completed five running trials (4.0 m/s ± 10%). A 10-camera system (200 Hz) captured marker positions and a force plate (1000 Hz) identified initial contact and toe-off events. A custom-written software program was used to filter the data, perform rigid body transformation procedures, and calculate angular velocities. Initial contact and peak angular velocity during three stance subphases were used for subsequent analysis. Two-way ANOVAs and Friedman's two-way ANOVAs for ranks were performed to investigate group and gender differences for normally and non-normally distributed dependent variables, respectively ($\alpha = 0.05$).

RESULTS

There were significant group main effects at initial contact and during all three stance subphases. At initial contact, PHP group medial midfoot eversion angular velocity was increased. Within the subphases, the PHP group lateral midfoot exhibited increased peak inversion angular velocity during early- and mid- stance and decreased angular velocity during propulsion (Tables 1-2). There were also significant gender main effects at initial contact and during the early stance and propulsion subphases. At initial contact, male participants' lateral midfoot external rotation angular velocity was increased. Female participants demonstrated increased peak rearfoot plantar flexion angular velocity during early stance and peak rearfoot eversion during propulsion (Tables 1-2).

Table 1: Initial Contact and Peak Early-Stance Angular Velocity (°/s) Mean (SD)

		Initial Contact		**Early-Stance**	
Variable	**Gender**	**CON**	**PHP**	**CON**	**PHP**
LMF-Tran	Female	*-22.7 (51)*[b]	*10.2 (86.6)*	-108 (24.4)	-9.8 (151.7)
	Male	*-57.6 (43.6)*	*-51.3 (80.8)*	-92.3 (28)	-132.6 (95.1)
MMF-Fron	Female	**-2.3 (33.9)**[a]	**-43 (50.2)**	22.9 (64.5)	-32.7 (88.5)
	Male	**-6.5 (46.5)**	**-39.9 (46.4)**	5 (54.6)	-49.5 (76.3)
LMF-Fron	Female	2.9 (82.4)	-5.5 (73)	**68 (82.6)**[a]	**85.7 (94.4)**
	Male	24.7 (33.9)	-4.1 (35.3)	**36.3 (65)**	**151.2 (48.6)**
RC-Sag	Female	-149.5(50.3)	-164.4 (125.5)	*-141 (148.3)*[b]	*-140.2 (191.5)*
	Male	-96.4 (84.4)	-121.6 (117.7)	*-25.3 (159)*	*-31.7 (178)*

LMF: Lateral Midfoot; MMF: Medial Midfoot; RC: Rearfoot Complex; [a]ANOVA; [b]Friedman's; Bold: Significant Group main effect; Italicized: Significant Gender main effect.

Table 2: Peak Mid-Stance and Propulsion Angular Velocity (°/s) Mean (SD)

		Mid-Stance		**Propulsion**	
Variable	**Gender**	**CON**	**PHP**	**CON**	**PHP**
LMF-Fron	Female	**9.7 (59.2)**[a]	**68.1 (59.7)**	**-22.8 (84.2)**	**-54.3 (75.6)**
	Male	**0.3 (59.1)**	**69.4 (107.8)**	**13 (55.7)**	**-97.7 (49)**
RC-Fron	Female	106.5 (44.4)	65.7 (78.5)	*-163.9 (24)*[a]	*-144.8 (70.4)*
	Male	10.7 (92.2)	119.1 (78.5)	*-115.4 (38.5)*	*-100.4 (17.6)*

DISCUSSION

The significant group differences were all in the frontal plane, and primarily involved the lateral midfoot. The frontal plane is a major plane associated with foot pronation, which is controlled largely by the plantar fascia and intrinsic foot muscles. Further, the lateral midfoot may be supported by fewer extrinsic dynamic stabilizers than the medial foot, and thus impacted to a greater extent when the plantar fascia is weakened. Finally, the significant gender main effects suggest there may be gender differences that should be considered when investigating joint angular velocity during running gait.

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DISCLOSURE STATEMENT

The authors have no conflicts of interest to disclose.

KINEMATICS AND MUSCULAR ACTIVATION WHILE PERFORMING ROUNDHOUSE KICK IN TAEKWONDO ELITE ATHLETE FROM MEXICO CITY

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INTRODUCTION

Taekwondo (TKD) is a combat sport which mostly uses kicks during the combat with the objective of getting points and eventually wining. Points are scored in TKD competition via kicks to the torso and to the head. In TKD it is important that athletes can perform their kicks with the higher possible force at the higher possible speed [1], [2]. As TKD is characterized by this higher speed and force kicks, it is also highly demanding for most muscle groups [3]. In addition, there are a variety of kicks styles within the sport, which are used to have different powers and effects. The most used kick in competition is the roundhouse kick (dollyo chagi), however, there exist different styles for performing one kick. The differences in the biomechanics of the individual kick styles can affect the results in the competition and may lead the athlete to an injury [4]. There are some studies focused on the biomechanics of frontal or roundhouse kicks [5][6], but there are only few that compare the difference between dominant and non-dominant leg of the subjects focused on the velocity [7], we believe that it is important to analyze the kinesiology of the dollyo chagi kick, comparing performance in both sides of the athlete to improve competitiveness and safety within the sport, also following the training sessions and helping in the competitive ranking of athletes.

CLINICAL SIGNIFICANCE

It is known that TKD athletes tend to favor one side when kicking, however, being able to kick with the same efficiency with both sides would give an advantage over the competition and also would help decrease any possible lesion due to imbalance between sides.

METHODS

To prove the importance of this, we took the data from one female subject part of Mexico's City Taekwondo Association (capable of taking part of the Pan-American Games), the subject has 16 years old, a height of 1.64 m, and weight of 53.9 kg. She performed 6 times the dollyochagi kick with each foot, aimed at a Digital Force Sensor (PunchSensor, USA) (DFS) located at the trunk and head height, the kick started with each foot on an Optima Force Plate (AMTI, USA), the subject had 14 mm reflective markers placed following the Plug-in Gait Model with additional markers at front of the tight and shank for better rotation calculations, also had Tringo IM sEMG sensor (Delsys, USA) bilateral at biceps femoris, semitendinosus, gluteus maximus and medius, rectus femoris and tensor fascia latae. When performing the kicks subject was instructed to hit as faster and hardest as she could, all data was captured in Vicon Nexus® (Oxford, UK) which has 9 Vero and 6 Vantage cameras with a capture rate at 200 Hz. All data was processed in Nexus and later in Visual3D (C-Motion, USA), results from the force plates were normalized to subjects' weight, while sEMG was normalized to the maximum voluntary contraction.

RESULTS

The results shown in this paper are for the leg that is kicking, in Figure 1 the angle/angle graph at knee and hip. On Table 1 the maximum muscle activation while performing the kick is shown.

For the force the maximum peak before kicking for the right leg was 2.6±0.2 while for the left was 2.3 ±0.1; the force received to the DFS was for the right side 2.85 and left side 3.14; hip range of motion (RoM) for the right leg was 40.79±3.79 while left 40.54±2.56; knee RoM was for the right leg 104.84°±7.17 and left was 93.28°±3.17; ankle RoM for the right leg was 76.58°±7.90 and left leg 46.81°±6.67.

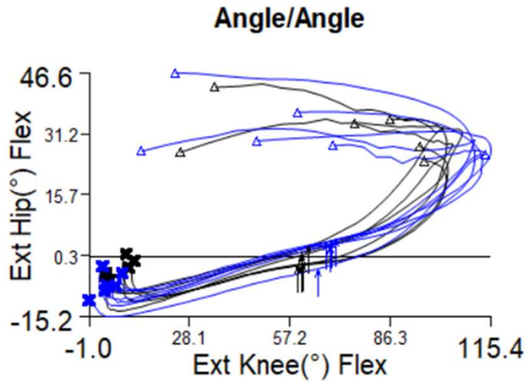

Figure 1: Angle/angle graph of knee and hip sagittal plane while performing dollyochagi kick.
Blue line show kicks performed with right leg, black line kicks performed with left leg. Cross show beginning of the kick, up arrow foot take off, and triangle moment of kick.

Table 1: Maximum Muscle Activation % while performing dollyochagi kick

Muscles	**Rectus F.**	**Biceps F.**	**Semitendinous**	**G. Maximus**	**G. Medius**	**Tensor**
Right	50 ±13	43± 7	81 ±19	52± 10	26 ±7	74± 12
Left	45± 16	176 ±15	124± 20	55± 15	54 ±10	127± 13

DISCUSSION

As the angle-angle graphs show in Figure 1, both legs follow a similar RoM for Hip and Knee when kicking, but we see that there exist asymmetries in the muscle activation specially in the hamstring muscles, gluteus medius and tensor, where the left side has a bigger % activation compared to the right side. We can conclude that the subject could work more on the right side based on the lower force generated at take-off and when kicking, and also by the muscle activation. This prove the importance of this type of methodology, hence with this information the coach or medical team of the athlete could create a plan to reduce the muscular imbalance, also, if both legs are balance and the athlete feels secure and safe to perform the kick with both legs, it could give an advantage since biomechanically and also physically, they could perform with either leg if the opportunity presents itself.

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DISCLOSURE STATEMENT

Authors declare that there is no conflict of interest.

THE MECHANICAL POWER THAT ELITE TAEKWONDO ATHLETES GENERATE WHILE PERFORMING DIFFERENT KICKS

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INTRODUCTION

Taekwondo (TKD) is a combat sport which mostly uses kicks during the combat with the objective of scoring points via kicks to the torso and to the head.

In TKD it is important that athletes can perform their kicks with the higher possible force at the higher possible speed (1). For that reason, mechanical power, which is expressed as the product of force and velocity, is an important factor in sports that require explosive energy such as TKD (2), because it provides combined information about how strong and how fast a kick hits.

CLINICAL SIGNIFICANCE

Being able to calculate the mechanical power of a kick while performing it, could give the trainers and the athlete more information regarding the effectiveness of their kick, which is different to the mechanical power often calculated while performing jumps, this result could give a bigger insight to the kick, and could help plan future goals for the athletes.

METHODS

We evaluate five (3 men and 2 women) TKD elite athletes (able to qualify to the PanAmerican Games) from Mexico City Sports Institute, aged 18.42±1.19 years old with an average height of 1.72±0.06m and an average weight of 61.88±7.98kg. The test consisted in the execution of five different TKD kicks (6 times each): cut (or frontal), dollyo chagui and pi chagui, which were aimed at the head and at the torso (intending a subject of 1.75 m height), while the chigo chagui was only aimed at the head, subjects were instructed to kick the fastest and strongest they could once a sensor turn on.

For this study we used the following equipment: two Optima-HPS force plates (AMTI®, Watertown, USA), two Digital Force Sensor (Punch Sensor®, Fremont, USA) mounted on a moving aluminum structure to measure the force of impact of every kick, these sensors have padding in order to avoid any injury. And a photogrammetry system (Vicon Oxford Metrics®, Oxford, UK), consisted of 15 infrared sensitive cameras (6 Vantage and 8 Vero), sampled at a rate of 200 Hz used to track the movement trajectories of 14 mm retroreflective markers place following the Plug-in Gait Full Body model.

Kinematic data was processed in Nexus (Vicon Oxford Metrics®, Oxford, UK) and all data was then exported to Visual 3D (C-Motion, Washington DC, USA).

A MATLAB (The MathWorks, Inc. Natick, Massachusetts, USA) code was designed to merge the data from the force plates, the kinematics, and the Punch Sensors where the mechanical power was calculated using the expression: $P[W]=F[N]*V[m/s]$ being F the force measured by the Punch Sensors and V the velocity of the toe of the kicking leg when it touches the Punch Sensor. This velocity was obtained by calculating the first derivative of the position of the toe marker. The forces were normalized using each subject weight. The velocity was normalized dividing the whole signal by the greatest velocity of the foot during its whole trajectory (3); since velocity and force values where normalized, the power is presented in normalized values.

RESULTS

The results are presented in means of all values calculated and measured.

In (Table 1) there is a comparison of power between kicks aimed to head or to the body. In (Table 2) there are all the values considered for the assessment of kicks with normalization.

Table 1: Mechanical power comparing head and torso kicks.

Kick	Power (W)	N. Power
Cut Head	3510.48±2381.85	0.79±0.446
Cut Body	4594.30±4525.35	1.00±0.855
Dollyo Head	9964.30±7068.94	1.38±0.808
Pi Body	13035.92±7880.92	1.64±1.141

Table 2: Normalized kinetics calculated from 3 different TKD kicks at impact.

Kick	Side	Exec. Time (s)	Force	Velocity	Punch Sensor Force	Power
Cut	R	0.27±0.099	0.42±0.242	0.44±0.227	2.13±0.450	0.92±0.599
	L	0.29±0.086	0.35±0.153	0.41±0.222	2.16±0.495	0.88±0.703
Dolpi	R	0.25±0.033	1.46±0.379	0.62±0.322	2.58±1.163	1.70±1.121
	L	0.24±0.026	1.44±0.298	0.54±0.255	2.54±0.951	1.32±0.829
Chygo	R	0.35±0.090	1.67±0.369	0.32±0.162	2.13±0.409	0.72±0.462
	L	0.44±0.109	1.56±0.337	0.19±0.14	1.72±0.489	0.37±0.357

DISCUSSION

There is no apparent correlation between the ground reaction force and the force received at the kick. Based on our results the lateral kicks are the ones with the higher power, being almost twice as the power from frontal kicks and almost three times the power from descendent kicks.

When analyzing the aim of the kick height, the kicks that were aimed to the head, had a lower power, compared to the ones aimed to the body. When the target is higher, they hit with a much lower force that could be explained by the effort required to elevate the leg.

Finally, frontal kicks are not apparently affected by the laterality of the athlete, since there is almost no difference between the values obtained with the right and left leg, contrary to what happens with the lateral and descendent kicks.

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DISCLOSURE STATEMENT

The authors have no conflicts of interest to disclose.

Quality Assurance Practices in Clinical Gait Analysis

Instructors: Eric Dugan, PhD; Amy Barbuto, PT, DPT, PCS; Jason Long, PhD; Alison Hanson, PT, MSPT, OCS

Purpose: The purpose of this course is to discuss quality assurance practices currently utilized by three different accredited motion analysis laboratories for clinical gait analysis.

Intended audience: All personnel who are involved in data collection, data processing, data interpretation, and/or treatment decision making as part of clinical gait analysis. This includes, but is not limited to, biomechanists, kinesiologists, engineers, physical therapists, technicians, and physicians.

Prerequisite knowledge: Participants should have the basic knowledge and familiarity of the clinical gait analysis process.

Outline of tutorial content:

In order to achieve the purpose of the tutorial, personnel from three accredited motion analysis laboratories will present on selected topics related to quality assurance practices. The first group will discuss how competencies for motion lab personnel are established, including biomechanists, physical therapists, technicians, and physicians. The second group will discuss reliability processes for marker placement, physical exam measures, and data processing. The third group will discuss processes for validation and determining reliability of technical systems, including cameras, force plates, and EMG.

Introduction (5-minutes): Eric Dugan

Establishing competency for motion lab personnel (20-minutes): Amy Barbuto, Eric Dugan

Marker placement, physical exam, and data processing reliability (20-minutes): Alison Hanson

Technical system validation & reliability (20-minutes): Jason Long

Question & Answer Discussion (25-minutes)

At the completion of this course, participants should be able to:

Understand core competency areas critical to motion lab personnel working in clinical gait analysis.

Discuss best practices for determining reliability for marker placement, physical exam measures, and data processing for clinical gait analysis.

Understand how technical systems are validated and reliability established for use in clinical gait analysis.

DOES COMPUTERIZED GAIT ANALYSIS ADD TO THE COST OF CARE FOR CHILDREN WITH CEREBRAL PALSY? AN UPDATE

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INTRODUCTION

Children with cerebral palsy (CP) have gait patterns that often involve deviations at multiple levels of the lower extremity. Computerized gait analysis (CGA) has enhanced our understanding of gait pathology in children with CP and allowed for a multilevel surgical approach (SEMLS) instead of sequential surgery (SSA). It is not currently known if CGA increases the overall cost of providing orthopedic surgical intervention for children with CP. Therefore, the purpose of this study was to determine the differences in billable provider charges for SEMLS based on CGA and SSA without CGA for orthopedic surgical treatment of ambulatory children with CP.

CLINICAL SIGNIFICANCE

Since the cost of CGA can be a barrier to receiving SEMLS, it is important to understand the total cost of the care package with CGA compared to the cost of the care package without CGA. If CGA does not add to the cost of care, more patients may be able to benefit from CGA and SEMLS as part of their orthopedic surgical care.

MATERIALS/METHODS

Charges associated with nine common orthopedic surgical combinations (both unilateral and bilateral, soft tissue or soft tissue plus bony) for children with CP were determined. The same nine combinations of procedures were then subdivided into SSA with either 1 or 2 surgical procedures per admission. The charges for each surgical combination were estimated based on current billing procedures and for associated CPT codes. Charges included surgery, anesthesia, operating room, recovery room, hospital stay, physical therapy (PT) and, for the SEMLS patients only, pre-operative CGA. Professional coders determined all surgical charges including multiple procedure charge reduction based on the Medicaid/Medicare model of billing.

Instrumented gait analysis included 3D kinematics and kinetics, surface electromyography, foot pressures, and physician interpretation. An additional charge was included for a fine wire EMG procedure for where posterior tibialis decisions were being considered. SEMLS and SSA costs across the 9 surgical combinations were compared using paired t-tests. The relationship between the total charge of SEMLS and the difference between SSA and SEMLS charges was examined using Pearson's correlations.

RESULTS

The total charge to complete each combination was higher for SSA than for SEMLS in all cases (Figure 1). The differential ranged from \$10,247 to \$75,069 with the percentage difference ranging from 20-47% (Table 1) which is more than the cost of CGA (\$3,513 - \$4,452). The mean difference between SEMLS and SSA was \$43,606 ($p=0.0002$). The dollar difference ($r=0.98$, $p<0.0001$) and percentage difference ($r=0.79$, $p=0.01$) were both related to the total charge of the SEMLS surgery.

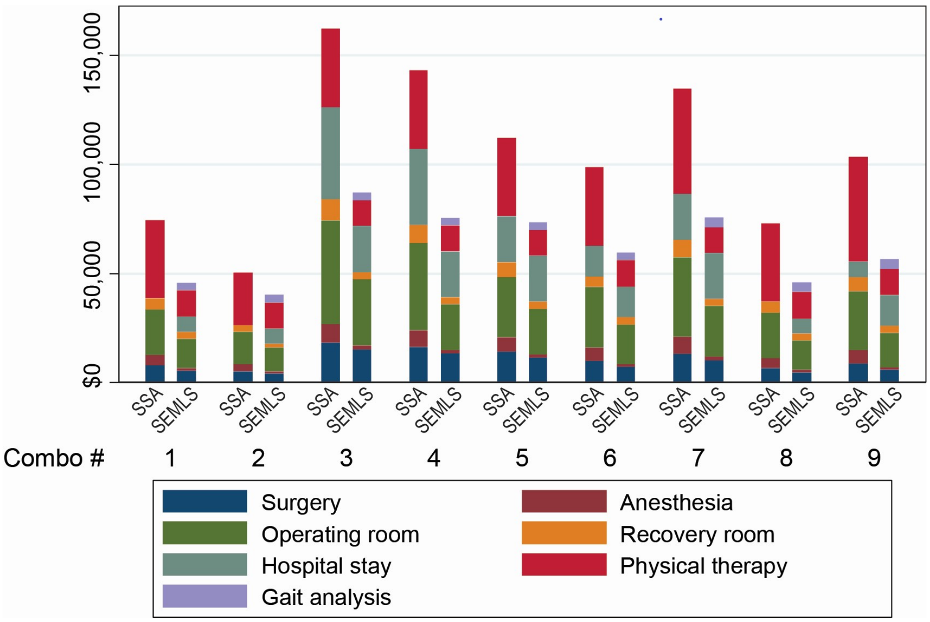

Figure 1: Charge of each surgical combination under SSA and SEMLS

Table 1: Charge difference between SEMLS and SSA for each surgical combination

Surgical Combination	Total # procedures	# admiss for SSA	SEMLS total	SSA total	Difference	% difference
1	6	3	$45,777	$74,448	$28,672	38.5%
2	4	2	$40,151	$50,398	$10,247	20.3%
3	12	3	$87,205	$162,273	$75,069	46.3%
4	10	3	$75,560	$143,262	$67,702	47.3%
5	8	3	$73,556	$112,157	$38,601	34.4%
6	8	3	$59,543	$98,820	$39,276	39.7%
7	7	4	$75,779	$134,684	$58,905	43.7%
8	4	3	$45,877	$72,942	$27,065	37.1%
9	5	4	$56,553	$103,469	$46,916	45.3%

DISCUSSION

This analysis shows that there are lower financial costs associated with SEMLS including CGA vs. SSA without CGA for the treatment of multilevel gait problems in children with CP. For every one of the 9 surgical combinations studied, the charges were higher for SSA than for SEMLS, regardless of the extent of surgery performed. Based on this finding, in addition to the other benefits of SEMLS, SEMLS including CGA is recommended over SSA for the treatment of gait disorders in children with CP. Other benefits of SEMLS include reduced anesthesia risks, reduced time away from school (for patients) and work (for caregivers), as well as improved understanding of gait pathology [1, 2] and improved treatment outcomes [3, 4] afforded by CGA.

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

GAIT IMPROVEMENT IN PATIENTS WITH KNEE OSTEOARTHRITIS AFTER PROXIMAL FIBULAR OSTEOTOMY

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INTRODUCTION

Knee Osteoarthritis (KOA) is a degenerative disease, which causes pain and disability. Since the medial compartment of the knee joint bears 60% to 80% of the load during gait, medial compartment KOA is commonly observed by knee varus deformities [1]. In recent years, proximal fibula osteotomy (PFO) has become a new choice for the management of medial compartment KOA, which has advantages of a smaller surgical incision, local anesthetic operation, shorter hospital stays and lower treatment cost [2]. However, the change of gait pattern after PFO to treat medial compartment KOA is still largely unknown. Therefore, the aim of this study was to evaluate the gait alterations in patients of medial compartment KOA after PFO.

CLINICAL SIGNIFICANCE

These findings provided evidence of a biomechanical benefit and gait improvement following PFO to treat medial compartment KOA which may apply to guide gait training and rehabilitation interventions for KOA patients after PFO.

METHODS

Nine females with medial compartment KOA (age 65.7±3.5 years, height 1.57±0.05 m, weight 62.5±6.7 kg) and nine healthy females (age 65.2±2.3 years, height 1.57±0.03 m, weight 62.8±7.5 kg) were recruited to form the KOA and control groups separately. Gait data with self-selected speed were collected using a 10-camera motion capture system and two force plates before and at 6 months after PFO for KOA group. Institutional review board approval was obtained and all participants provided written informed consent.

Paired t-test and independent t-test were performed to determine the effect of PFO within the KOA group and also between KOA and control groups for spatiotemporal, kinematic and kinetic variables. A significance level of 0.05 was used for comparisons.

RESULTS

The single support and swing phases were significantly different between the affected and unaffected limbs in the KOA group preoperatively, which was not observed postoperatively (Table 1). KOA patients still showed a decreased step length, step cadence, single support phase and swing phase, but increased step time before and after PFO compared to controls. The knee peak flexion and sagittal range of motion (RoM) in the affected limb were smaller than unaffected limb preoperatively, which became similar between limbs postoperatively. Additionally, the frontal RoM of hip and knee joints significantly decreased in the affected limb after PFO. A significantly different peak anteroposterior ground reaction force, external adduction moments for hip and knee joints between limbs preoperatively disappeared after PFO at 6 months.

Table 1 The parameters expressed as mean (SD) for KOA and control groups

	Preoperation		Postoperation		Control
	affected	unaffected	affected	unaffected	
Single support phase (%cycle)	31.23(7.28)*	34.89(5.64)*#	35.30(3.80)*	36.17(3.44)*	39.15(3.80)
Swing phase (%cycle)	36.44(3.94)*	32.78(5.39)*#	35.92(3.72)*	34.66(3.66)*	38.94(8.18)
Hip frontal RoM (degree)	8.80(2.74)*	7.14(1.66)*	7.29(2.13)*#	7.11(0.99)*	10.99(1.03)
Knee peak flexion (degree)	27.41(19.03)*	44.03(10.58)*#	42.75(17.08)*#	42.72(14.14)*	57.69(4.73)
Knee sagittal RoM (degree)	26.40(13.48)*	43.62(8.39)*#	39.08(15.47)*#	44.42(11.84)*	58.53(4.24)
Knee frontal RoM (degree)	25.33(14.41)	17.77(7.67)	13.80(9.62)#	16.09(7.50)	16.28(4.24)
Hip adduction moment (Nm/kg)	0.71(0.22)*	0.83(0.20)*#	0.88(0.24)#	0.97(0.15)	0.97(0.09)
Knee adduction moment (Nm/kg)	0.55(0.16)	0.67(0.11)*#	0.59(0.15)	0.70(0.13)*	0.49(0.14)
Anterior GRF (N/kg)	1.11(0.48)*	1.30(0.37)*#	1.22(0.44)*	1.19(0.42)*	1.71(0.18)
Posterior GRF (N/kg)	1.15(0.34)*	1.28(0.45)*#	1.12(0.32)*	1.27(0.45)*	1.61(0.18)

* Significant difference compared with controls
# Significant difference compared with affected limb preoperatively

DISCUSSION

Improved hip and knee joint functions in the affected limb after PFO were verified by knee peak flexion, peak external HAM, knee sagittal and frontal RoMs. Moreover, gait symmetry improved postoperatively and was confirmed by single support and swing phases, knee peak flexion and sagittal RoM, peak external HAM and KAM, peak anterior and posterior GRF. The phenomenon of knee pain relief postoperatively seemed to not cause a reduction of medial knee loading, but a redistributed knee pressure by a more neutral alignment. The present results provided biomechanical evidence of the feasibility and effectiveness of PFO to treat medial compartment KOA although a longer follow-up period was still needed to better understand the potential benefits and limitations of PFO

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DISCLOSURE STATEMENT

There are no conflicts of interest to disclose.

A multicenter study classifying sagittal plane gait deviations of children with cerebral palsy into clinically meaningful biomechanical phenotypes using a novel algorithm

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INTRODUCTION

Identifying the gait pattern used by patients with cerebral palsy (CP) is a useful part of the treatment decision making process.[1-4] Clinical gait analysis provides quantitative information that can be used to quantitatively categorize gait patterns.[5] The objective of this work was to report on the prevalence of clinically meaningful sagittal plane gait phenotypes in children with CP across four motion analysis centers in the United States using a novel algorithm.

CLINICAL SIGNIFICANCE

Deriving these clinically meaningful biomechanical phenotypes can provide objective classifications helpful in prescribing interventions.

METHODS

Data was used from 700 subjects that had completed instrumented gait analysis between January 1, 2019 and December 31, 2020. Discrete kinematic variables for the hip, knee, and ankle were extracted from a shared registry. Classification categories were developed based on definitions from literature and physician review (Figure 1). Sagittal classifications included crouch, recurvatum, true and apparent equinus. Outlier data were removed based on averaged motion parameters collected from typically developing children at the 4 participating labs (Table 1).

Table 1: Cut-off values for gait parameters defining sagittal plane classifications

Gait Pattern	Peak Stance Ankle Dorsiflexion	Mean Stance Knee Flexion	Min Stance Knee Flexion	Min Hip Flexion	Knee Range of Motion
Apparent Equinus	> 0° to ≤ 4.4°	> 23.1°	---	> 6.9°	---
True Equinus	≤ 0°	> 23.1°	---	> 6.9°	---
Recurvaturm	---	---	≤ 0°	---	---
Crouch	> 20.3°	> 23.1°	---	> 6.9°	---
Stiff Knee	> 4.4° to ≤ 20.3°	> 4.4° to 23.1°	---	≤ 6.9°	≤ 43.3°
Normal	> 4.4° to ≤ 20.3°	> 4.4° to 23.1°	---	≤ 6.9°	> 43.3°

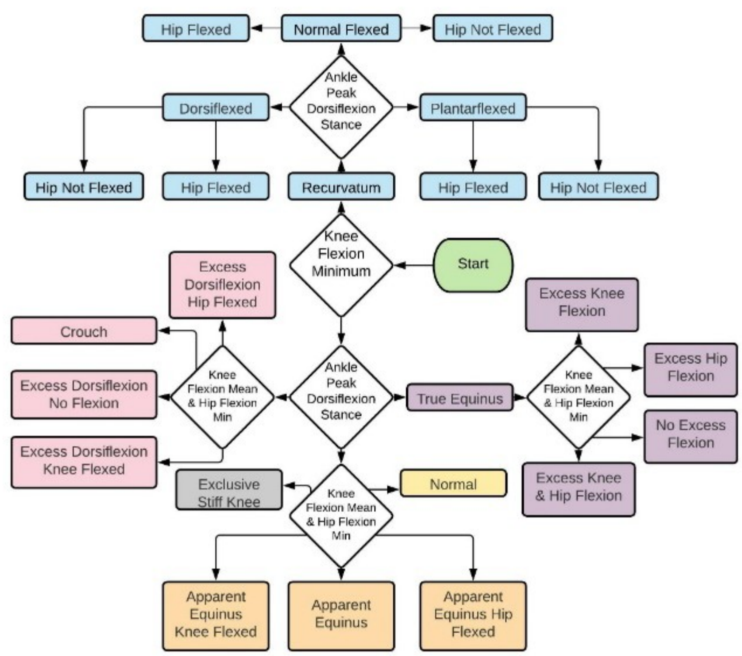

Figure 1: Algorithm to classify sagittal plane gait patterns

RESULTS

Normal was the most commonly occurring sagittal phenotype for GMFCS I (41.3%) and II (26.9%). However, this total may have been influenced by inclusion of the non-involved side of children with hemiplegia. Knee recurvatum with normal ankle and no excess hip flexion was the most common sagittal deviation in GMFCS I. Excess ankle dorsiflexion was the most common sagittal deviation in GMFCS Level II (17.9%), III (31.6%), and IV (27.8%). The distribution of all biomechanical phenotypes is included in Table 2.

DISCUSSION

Deriving clinically meaningful biomechanical phenotypes can provide objective classifications to prescribe interventions. This novel algorithm could allow for automating the classification process into which would lead to improved efficiency and repeatability. While excess ankle dorsiflexion was the most common sagittal deviation in GMFCS II, this may be due to a single segment foot model being unable to register equinus associated with a midfoot break. This is not a shortcoming of the algorithm necessarily however a segmental foot model may be needed to properly classify foot gait deviations. This work demonstrates the ability of the algorithm to be applied across multiple centers. Future work aims to expand this algorithm to other joint measures, and to refine the algorithm beyond use of discrete data points.

Table 2: Prevalence of sagittal plane classifications by GMFCS level

Sagittal Plane Deviation	GMFCS I n=1491	GMFCS II n=2057	GMFCS III n=892	GMFCS IV n=90	Total Population
Percent of Total Population	**32.5%**	**44.8%**	**19.4%**	**2.0%**	**100.0%**
Apparent Equinus Multi-Level Flexion	0.7%	0.9%	1.3%	3.3%	1.0%
Apparent Equinus Knee Flexion	0.1%	0.4%	1.0%	0.0%	0.4%
Apparent Equinus Hip Flexion	0.3%	0.4%	0.9%	0.0%	0.5%
Apparent Equinus No Flexion	1.7%	2.7%	0.9%	0.0%	2.0%
Apparent Equinus Total	**2.8%**	**4.4%**	**4.1%**	**3.3%**	**3.8%**
Knee and Hip Flexion	4.4%	5.3%	16.1%	23.3%	7.6%
Only Hip Flexion	5.6%	6.7%	9.9%	8.9%	7.0%
Only Knee Flexion	5.0%	5.3%	5.5%	4.4%	5.1%
Exclusive Stiff Knee	4.8%	4.2%	2.8%	1.1%	4.0%
Crouch Multi-Level Flexion	2.5%	7.0%	20.2%	12.2%	8.2%
Excess Dorsiflexion Hip Flexion	0.7%	2.2%	2.5%	4.4%	1.8%
Excess Dorsiflexion Knee Flexion	2.5%	4.3%	6.3%	7.8%	4.2%
Excess Dorsiflexion No Flexion	3.6%	4.5%	2.7%	3.3%	3.8%
Excess Dorsiflexion Total	**9.5%**	**17.9%**	**31.6%**	**27.8%**	**18.1%**
Recurvatum with Plantar Ankle & No Excess Hip Flexion	4.0%	5.2%	5.3%	1.1%	4.7%
Recurvatum with Plantar Ankle & Excess Hip Flexion	0.2%	0.9%	2.0%	12.2%	1.1%
Recurvatum with Normal Ankle & No Excess Hip Flexion	15.3%	13.3%	3.7%	0.0%	11.9%
Recurvatum with Normal Ankle & Excess Hip Flexion	1.3%	1.2%	1.6%	0.0%	1.3%
Recurvatum with Dorsi Ankle & No Excess Hip Flexion	0.6%	0.4%	0.6%	0.0%	0.5%
Recurvatum with Dorsi Ankle & Excess Hip Flexion	0.0%	0.0%	0.0%	0.0%	0.0%
True Equinus Multi-Level Flexion	1.1%	4.0%	5.0%	8.9%	3.3%
True Equinus Knee Flexion	1.1%	1.0%	1.8%	1.1%	1.2%
True Equinus Hip Flexion	0.3%	0.9%	2.0%	0.0%	0.9%
True Equinus No Flexion	2.8%	2.4%	1.9%	1.1%	2.5%
True Equinus Total	**5.4%**	**8.2%**	**10.8%**	**11.1%**	**7.9%**
Normal	41.3%	26.9%	6.1%	6.7%	27.2%

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DISCLOSURE STATEMENT

Dr. Davids is a consultant to Orthopediatrics Inc. All other authors have nothing to disclose.

EFFECTS OF A MEDIAL UNLOADER BRACE ON GAIT MECHANICS IN PATIENTS WITH OSTEOCHONDRITIS DISSECANS

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INTRODUCTION

Osteochondritis dissecans (OCD) of the knee occurs when an area of subchondral bone sustains an avascular insult and may begin to separate from the surrounding area [1]. Males typically present with an OCD lesion more often than females, and 66% of lesions are located in the medial femoral condyle [2]. OCD lesions can be treated operatively or with conservative treatment which traditionally includes casting, bracing, and/or activity restrictions. Specifically, an unloader brace applies forces above and below the knee joint to reduce the contact forces at the location of the OCD lesion in the knee. This treatment option may result in improved compliance due to the patient's ability to continue daily activities while maintaining brace wear. Secondly, the unloader brace may have biomechanical advantages for OCD lesion healing. However, the biomechanical efficacy of a medial unloader brace has not been confirmed in a pediatric population.

CLINICAL SIGNIFICANCE

To determine whether the use of a medial unloader knee brace alters gait mechanics in patients with medial femoral condylar OCD. If a reduction in valgus moment is observed, there may be biomechanical advantages to wearing this brace for patients with a medial OCD.

METHODS

Fifteen medial femoral condylar OCD patients (14 males, 12.7 ± 2.6 years, 156.9 ± 14.8 cm, and 52.0 ± 16.2 kg) were tested wearing a medial Ossur Unloader One knee brace. All patients performed over-ground walking at a self-selected speed under three conditions: no brace (NB), unadjusted (U) brace tension, and adjusted (A) brace tension as prescribed by their orthotist. A 14-camera motion capture system (Vicon Motion System Ltd, Denver, CO, USA) with 4 AMTI forceplates was used to collect kinematic and kinetic data at 120 Hz and 2880 Hz, respectively. Retroreflective markers were placed on specific bony landmarks and a custom Matlab (Matlab 2020b, Natick, MA, USA) model was used to compute joint angles and moments for the hip and knee. Wilcoxon signed-rank tests were performed to identify statistical differences between brace conditions (α=0.05).

RESULTS

Between braced conditions, peak coronal hip angle and mean coronal knee angle were not statistically significant (Table 1). However, peak internal knee valgus moment consistently decreased with each brace condition (NB-A: 0.06 Nm/kg, p=0.017). Similarly, hip abduction moment at the second peak of SLS decreased while wearing the brace (NB-U: p=0.061, NB-A: 0.07 Nm/kg, p=0.009) (Figure 1).

Table 1. Coronal plane hip and knee kinematics and kinetics (Mean ± SD) for each brace condition.

Variable	No Brace	Unadjusted	Adjusted
Single-Limb Stance Phase			
Hip Ab/Adduction Max. (˚)	3.6 (4.1)	2.6 (3.3)	3.1 (3.6)
Knee Valgus Avg. (°)	3.4 (3.5)	3.3 (3.6)	3.1 (3.6)
Knee Valgus Mom. Max. (Nm/kg)	0.37 (0.16)	0.35 (0.19)	**0.31 (0.20)** [N]
Single-Limb Stance Phase – Second Peak			
Hip Moment Max. (Nm/kg)	0.58 (0.13)	0.52 (0.11)	**0.51 (0.14)** [N]

Note: Significant differences from the No Brace condition notated in bold with a superscript N.

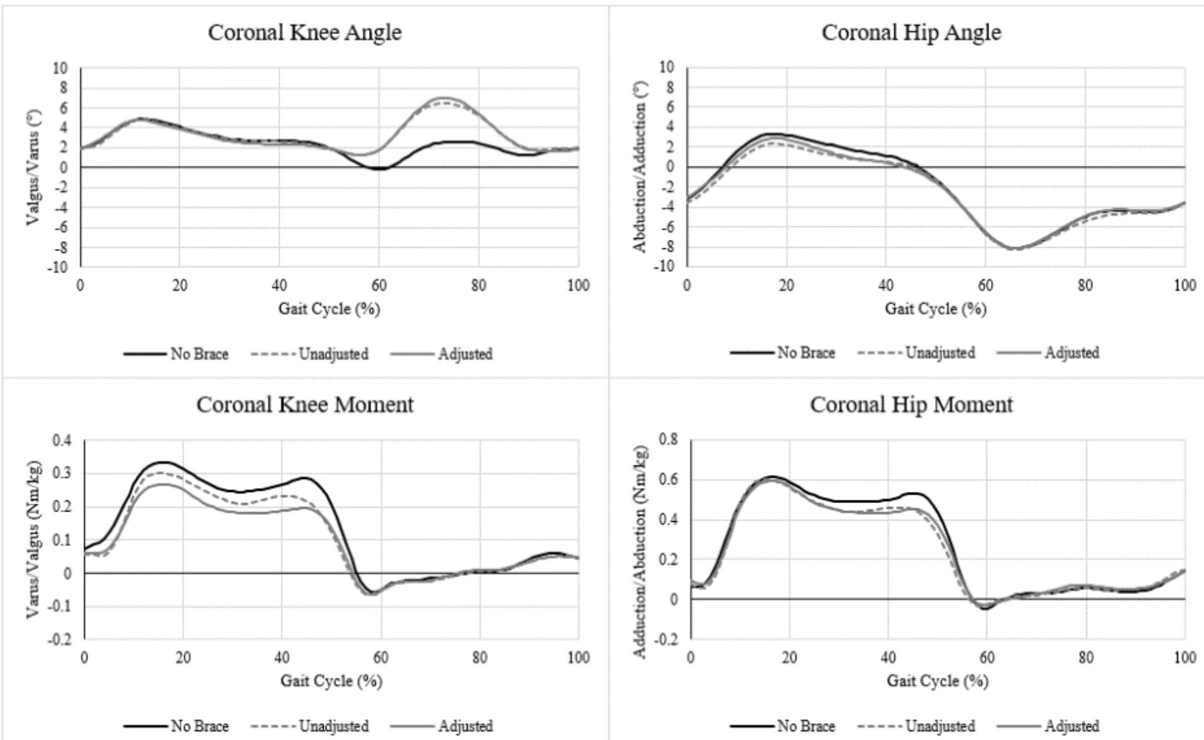

Figure 1. Average coronal hip and knee angle and moment throughout gait cycle

DISCUSSION

No significant differences were seen in hip or knee kinematics during single limb stance. Although knee alignment did not change when wearing the brace, peak knee valgus moment was decreased during walking with an adjusted medial unloader brace. These results highlight potential biomechanical advantages of a medial unloader brace for medial femoral condylar OCD. Future work will assess the effects of compliance on patient outcomes and compare different unloader brace designs.

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MULTI-SEGMENT FOOT JOINT COORDINATION VARIABILITY IN YOUNG ADULT RUNNERS WITH PLANTAR HEEL PAIN

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INTRODUCTION

Plantar Heel Pain (PHP) is a common lower extremity injury experienced by recreational runners, with the highest prevalence in young adults [1]. Up to 20% of patients may continue to have pain 4 years after the initial onset of symptoms [2], which can have negative impacts on HRQOL and mobility [3]. Joint coordination variability during running may be an important factor in persistent PHP, however, the research to date has only investigated PHP during walking [4]. We hypothesized that young adults with PHP would show decreased joint coordination variability within the distal foot compared to healthy subjects during running.

CLINICAL SIGNIFICANCE

Dysfunctional coordination variability within the foot could be a factor in the mechanical overload of the plantar fascia and the development and/or persistence of PHP in runners.

METHODS

Twenty-six young adults participated. Plantar heel pain group participants (PHP) (7 F/6 M, Age: 29.0 ± 8.3 y., BMI: 23.1 ± 2.0 kg/m^2) had clinical signs and symptoms for at least six weeks, ran at least 10 mi/week. Healthy, uninjured participants (CON) (6 F/7 M, Age: 30.5 ± 5.9 y., BMI: 23.5 ± 3.1 kg/m^2) were sex, age (± 5 y.), BMI, and running mileage matched with the PHP group. Retroreflective markers on each participant's foot and shank defined seven segments and six functional articulations (rearfoot, medial and lateral midfoot, medial and lateral forefoot, and 1st metatarsophalangeal) [5]. Each participant performed 5 shod (sandal) overground running trials along a runway with an embedded force plate (1000 Hz), while marker position data were collected using a 10-camera motion analysis system (200 Hz). Joint coordination between segments of interest was calculated [6]. The variability during early stance, midstance, and propulsion stance subphases was used for subsequent analysis. Two-way ANOVAs for normally distributed dependent variables and Friedman's two-way ANOVAs for ranks for non-normally distributed dependent variables were performed to investigate group and sex differences in the coordination variables ($\alpha = 0.05$).

RESULTS

Significant group x sex interactions were revealed for the Rearfoot$_{Fron}$-Rearfoot$_{Tran}$ during the propulsion subphase, and for the Rearfoot$_{Fron}$-Medial Midfoot$_{Fron}$ during the midstance and propulsion subphases. However, follow-up tests were only significant for the Rearfoot$_{Fron}$-Medial Midfoot$_{Fron}$ during propulsion. During the subphase, PHP females demonstrated increased coordination variability versus CON females. Significant group main effects were revealed for the Rearfoot$_{Fron}$-Medial Midfoot$_{Fron}$, Rearfoot$_{Fron}$-Lateral Midfoot$_{Fron}$, and Lateral Midfoot$_{Fron}$-Lateral Forefoot$_{Sag}$. For all the group main effects, the PHP group demonstrated increased variability versus the CON group (Table 1).

Table 1: Joint Coordination Variability (°) Mean (SD) results.

Coordination Variables	CON		PHP	
	Female	Male	Female	Male
$Rearfoot_{Fron}$-$Rearfoot_{Tran}$				
Propulsion[1]	2.6 (1.2)	3.1 (1.2)	4.2 (1.6)	2.6 (0.9)
$Rearfoot_{Fron}$-Medial $Midfoot_{Fron}$				
Early-stance[2]	0.9 (0.7)	1.1 (0.6)	1.7 (1.1)	1.6 (0.8)
Mid-stance[1F]	17.3 (6.9)	11.7 (7.2)	9.3 (8.6)	15.7 (5.7)
Propulsion	2.4[3] (1.8)	4.9 (2.2)	4.9[3] (1.8)	3.9 (2.2)
$Rearfoot_{Fron}$-Lateral $Midfoot_{Fron}$				
Early-stance[2F]	9.5 (8.1)	10.3 (6.2)	15.6 (7.6)	17.0 (7.9)
Lateral $Midfoot_{Fron}$-Lateral $Forefoot_{Sag}$				
Propulsion[2]	3.4 (1.1)	3.5 (1.5)	4.9 (1.8)	4.6 (1.4)

[1/1F] Significant ANOVA/Friedman's group*sex interaction, follow-up independent t-tests/Mann-Whitney U tests were not significant ($p > 0.05$); [2/2F] Significant ANOVA/Friedman's group main effect ($p < 0.05$); [3] Significant ANOVA group*sex interaction, significant follow-up independent t-test ($p < 0.05$).

DISCUSSION

Although there were significant differences between runners with and without PHP, our initial hypothesis that runners with PHP would exhibit decreased coordination variability within the foot segments was not supported. Rather, when there were significant differences, the PHP group demonstrated increased variability. The increased variability may have been the result of decreased foot stability due to plantar fascia degeneration and/or intrinsic foot muscle weakness believed to be associated with PHP. Finally, the significant group x sex interaction for the $Rearfoot_{Fron}$-Medial $Midfoot_{Fron}$ during propulsion suggests that there may be some sex differences in the way that PHP affects distal foot function during running gait.

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DISCLOSURE STATEMENT

The authors have no conflicts of interest to disclose.

EFFECT OF DIFFERENT SEVERITY OF FLEXIBLE FLATFOOT ON WALKING FUNCTION IN SCHOOL-AGE CHILDREN

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INTRODUCTION

Flexible Flatfoot (FFF) is an important classification of Flat Foot (FF) and is mainly seen in school-aged children. In early infancy, FFF is common in children below 6 and the arch of foot rapidly develops between the ages of 2 and 6. If the arch has not well formed after the age of 6, it is considered as a pathological abnormality [1]. The incidence of flatfoot in children is increasing continually and now the flatfoot has become one of the most common complaints in pediatric orthopedics [2]. The most common symptoms are foot pain, fatigue and decreased ability to perform daily activities [3]. However, the effect on motor function caused by different severity of clubfoot in school-age children is not known.

The purpose of this study is to investigate the effects of different severity of flexible flatfoot on walking function in school-aged children and to provide a basis for the selection of effective interventions and appropriate timing of interventions at a later study.

CLINICAL SIGNIFICANCE

To investigate the effect of different severity of flexible flatfoot on walking function in school-age children.

METHODS

A total of 158 children aged 7-12 years with flexible flat feet and 30 children with normal feet were recruited. Children with flexible flatfoot were screened according to clinical flatfoot grading [4] and were classified into mild, moderate and severe groups. 30 cases (60 feet) were randomly selected in each group and the gait data was collected under walking condition by applying the 3D motion gait analysis. This study has been examined by the ethical review committee of Yueyang Hospital.

In this study, a modified conventional model[5] was used with a total of 24 markers. A 19-camera (Motion Analysis Corporation, Santa Rosa, CA) motion analysis system combined with 4 force plate (Bertec) was used to collect the motion data and ground force data. spatio-temporal parameters, kinematic and kinetic data were analysed using motion software Visual 3D version 6.01.36 (C-Motion Corporation). Subjects walked barefoot at free walk speed. About ten to fifteen gait cycles were averaged for each participant.

Statistical analysis was performed using SPSS version 20 (SPSS Inc., Chicago, USA). Independent samples t-test and KRUSKAL-Wallis H-test were used to analyse the data.

RESULTS

Compared with the normal children group, the step length, the step speed and the percentage of swing phase were significantly reduced, and the step width and the percentage of stance phase increased significantly ($p<0.01$) in the children with all flat feet groups. No significant difference between groups of children with flat feet. The ankle plantarflexion angle on initial contact in the mild group was significantly increased compared with the normal group ($p<0.01$) and the range of motion (ROM) of hip flexion/extension was significantly reduced ($p<0.01$). The maximum power of the hip joint was significantly reduced in the mild and moderate groups compared with the normal group (p<0.05). The ROM of ankle flexion/extension was significantly reduced in the severe group compared with the normal group (p<0.05). The angle of inversion of the ankle in the flatfoot groups were smaller than that in the normal group (p<0.01). And there was a significant difference between the moderate and severe groups(p=0.006).

Table 1: Comparison of spatio-temporal parameters (mean± SD)

Group	Step Length (cm)	Step Speed (cm/s)	Step Width (cm)	Stance Phase (%)	Swing Phase (%)
Mild	53.94±6.09	112.03±12.81	10.57±2.31	61.63±1.62	38.37±1.62
Moderate	53.26±5.76	110.00±10.84	10.43±2.25	61.59±1.55	38.41±1.55
Severe	54.44±5.01	115.71±11.14	10.54±2.12	61.59±1.73	38.41±1.73
Normal	56.73±5.40	119.03±11.53	9.21±1.69	60.42±1.26	39.58±1.26
F/Z	4.337	21.489	14.245	8.490	8.490
P	0.005*	0.000*	0.003*	0.000*	0.000*

Note:*indicates a significant difference between the flatfoot groups and the normal group. No significant difference between groups of children with flat feet.

Table 2: Comparison of kinematic and kinetic parameters (mean± SD)

Group	Ankle ROM (°)	Knee ROM (°)	Hip ROM (°)	Foot orientation (°)	Ankle inversion (°)	Ankle Power (W/kg•m)	Knee Power (W/kg•m)	Hip Power (W/kg•m)
Mild	28.95±5.23	60.64±5.51	40.68±5.18	-3.67±5.50	7.33±4.30	2.24±0.55	0.22±0.15	0.35±0.18
Moderate	29.04±4.65	60.91±4.95	41.08±4.83	-5.58±5.78	5.68±4.68	2.13±0.47	0.26±0.17	0.34±0.17
Severe	27.67±5.69	60.88±4.89	42.00±3.61	-5.23±8.27	8.45±3.87#	2.18±0.57	0.24±0.14	0.38±0.17
Normal	30.30±4.69	59.74±5.03	43.49±6.14	-8.29±5.86	10.55±2.33	2.07±0.44	0.22±0.14	0.40±0.13
F/Z	2.748	3.143	9.956	16.634	42.297	2.885	2.412	8.071
P	0.043*	0.370	0.019*	0.001*	0.000*	0.410	0.491	0.045*

Note:*indicates a significant difference between the flatfoot groups and the normal group. Significant differences between the moderate and severe groups are indicated by **#(** p<0.01**).**

DISCUSSION

The effect of flexible flatfoot on walking function in school-age children is significant. children with mild flatfoot showed more abnormalities in joint coordination than children with moderate or severe flatfoot. The effects of flexible flatfoot on walking function in school-age children did not show a progressive change with increasing severity.

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DISCLOSURE STATEMENT

The authors declare no conflicts of interest.

Pilot Study: Brace Evaluation in Children with Cerebral Palsy-A Case Series

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INTRODUCTION

Cerebral palsy (CP) is a non-progressive neurological disorder that results in a constellation of mobility difficulties and presents differently in each affected child. Children with CP are often prescribed Ankle-Foot Orthoses (AFOs) to correct gait deviations, prevent joint deformity and improve function. Standard AFOs (S-AFOS) are traditionally designed with the ankle joint set at a neutral (or 90-degree) angle. New theories[1] have emerged to suggest this may not appropriately allow for optimal segmental alignment. Rather, customizing the Angle of the Ankle in the AFO (AAFO) and the Shank-to-Vertical Angle (SVA) for each limb may result in improved gait parameters, and ultimately, serve to prevent age and growth related musculoskeletal deformities. Tuned-AFOs (T-AFOs) are fabricated such that the AAFO is set to accommodate the functional range of motion at the ankle joint without inducing deformation of the subtalar joint, compromising range of motion or alignment of structures above and/or below the ankle. Heel posting and other customizations are added to the T-AFO in an effort to manipulate the ground reaction forces (GRFs) at the knee and hip joints, thus providing more typical alignment and joint forces during braced walking. This study aimed to determine if a T-AFO can more effectively restore typical gait kinematics than an S-AFO for patients with CP, as defined by Perry's prerequisites for normal gait[2]. We hypothesize that the T-AFO would demonstrate gait measures more closely approximating those seen in typically developing controls.

CLINICAL SIGNIFICANCE

Children with CP often demonstrate muscle spasticity, atypical gastroc-soleus muscle range of motion, and decreased selective motor control; frequently resulting in multiplanar, compensatory motion at the ankle and knee joints. Absent heel-strike and crouched gait are common deviations for which AFOs are prescribed. S-AFOs often inadequately address gait deviations. Setting an AFO in a neutral (or 90 degree) ankle angle (AAFO) may take all, or most, of the available muscle range from the distal end of the spastic or shortened gastrocnemius muscle, therefore limiting the available range of motion at the proximal end (knee). The patient then walks with either initial contact at the forefoot or a flexed-knee gait pattern while using the S-AFO. The T-AFO has been observed to accommodate for the range of motion limitations at the ankle joint while allowing better knee joint excursion (usually extension) during standing and walking activities. This study aims to quantify these anecdotal findings using quantitative gait analysis.

METHODS

The inclusion criteria for this study was comprised of patients who were ambulatory and used previously prescribed standardly configured AFOs for daytime ambulation. The population for this IRB approved study included five children (ages 3.8-11 years; 4 female and 1 male) with confirmed CP (4 Gross Motor Functional Classification System (GMFCS) level II and one GMFCS level III). Each participant completed instrumented, three-dimensional gait analysis (3DGA) using their S-AFOs during an initial visit and one month following delivery of study-provided, customized T-AFOs along with standardized footwear. Gait data were collected at a self-selected walking speed using a 14-camera motion analysis system (Vicon, UK) with the Plug-in-Gait Model. Data was exported to a custom software for analysis. Data extracted for analysis included single leg stance time, sagittal plane range of motion (ROM) of the tibia throughout stance, step length (SL), knee flexion at initial contact (IC), contralateral (CL) hip flexion minimum ROM, ipsilateral (ISP) hip flexion maximum ROM, and sagittal tibial tilt at IC. Key components of the gait cycle were divided into the following, quantifiable elements: Stability in Stance (single leg stance time, SLS; sagittal plane ROM of tibia during stance; and SL), Increased Step Length (knee flexion at IC; CL hip flexion minimum; ISP hip flexion maximum), and Pre-Positioning of the Foot (sagittal tibial tilt at IC). Paired t-tests were used to compare S-AFO and T-AFO gait data.

RESULTS

In the S-AFO condition, 8/10 limbs were deficient in Stability in Stance, 9/10 were deficient in SL, 8/10 showed increased knee flexion at IC and 10/10 were deficient with ipsilateral hip flexion maximum and sagittal tibial tilt at IC. During walking in the T-AFOs + standard footwear, subjects demonstrated the following improvements (Table 1): increased Stability in Stance (90% of limbs), increased SL (50% of limbs), decreased knee flexion at IC (50% of limbs), decreased CL hip flexion minimum with decreased IPS hip flexion maximum (90% of limbs) and improved Pre-Positioning of Foot/Sagittal Tibial Tilt at IC (50% limbs).

Table 1: Summary of Standard and Tuned AFO variables of interest
*indicates statistically significant differences ($p<0.05$)

	Stability in Stance		Increased Step Length				Pre-positioning of Foot
(control average ± 1 SD)	SLS Time (38-42)	Sag Tib ROM Stance (71.4- 61.0)	Step Length (.52-.66)	Knee Flexion at IC (1.5-14.3)	CL Hip Flexion Min (-13.8 - -1.8)	IPS Hip Flexion Max (31.3-43.5)	Sag Tib Tilt at IC (-20, -15.4)
S-AFO Average (SD)	36.2 (5.75)	48.7 (8.84)	0.39 (0.08)	27.6 (10.53)	-1.04 (5.88)	52.4 (8.03)	4.14 (6.86)
T-AFO Average (SD)	34.8 (7.45)	54 (12.35)	0.41 (0.08)	22.7 (17.58)	-0.95 (8.13)	45.4 (5.38)	0.02 (12.26)
t-test	0.52	0.08	0.36	0.22	0.96	0.01*	0.13
Change from Standard to Tuned							
1L	-3	21	0.09	-21	-2.6	-10	-15
1R	4	21	0.06	-21	-6	-1	-16
2L	1	7	0	-4	7	-1	-10
2R	-6	1	0.05	7	5	6	0.8
3L	0	3	-0.01	-15	-7	-16	-8
3R	-2	-2	-0.01	-11	-2.5	-16	2
4L	6	-1	0.08	11	-2	-1	4
4R	8	1	0.07	1	0	-9	2
5L	-10	1	-0.03	-2	0	-10	-5
5R	-12	1	-0.11	6	9	-12	4
Red=worsened, orange=no change, green=improved							
% improve	50%	80%	50%	60%	20%	90%	50%
% worse	40%	10%	20%	40%	20%	10%	50%
% no change	10%	10%	30%	0%	60%	0%	0%

DISCUSSION

Differences between S-AFO vs. T-AFO walking, though not statistically significant are likely an effect of the small sample size. The improvements demonstrated during Tuned AFO walking may be considered clinically significant in that the changes moved the identified prerequisites of normal gait closer to those of typically developing children. Results of this pilot study, although limited by sample size and the heterogeneity of the population studied, demonstrate the potential benefits of walking in customized, T-AFOs for ambulatory children with CP. Further studies are needed to demonstrate greater quantifiable evidence of improved gait parameters during use of T-AFOs. Additionally, future investigation is required to help identify appropriate candidates for tuned orthotics.

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CONTINUOUS INTER-LIMB COORDINATION AND STABILITY IN VETERANS AND SERVICE MEMBERS WITH TRANSTIBIAL AMPUTATION

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INTRODUCTION

The number of Veterans and Service Members (SMs) with transtibial amputation (TTA) is growing due to the aging population with dysvascular disease and diabetes, as well as from U.S. military involvements abroad [1,2]. With high associated healthcare costs [3], it is expected that conclusive research is available to clinicians for proper prosthetic prescription. However, most research is noncommittal and lacks guidance for clinical practice [4]. Traditional biomechanical parameters indicate mixed results, limiting evidentiary support for optimal prescription guidelines [5]. Conversely, continuous measures of coordination (the movement relationship between limbs) and stability (the ability to offset a perturbation) derived from Relative Phase (RP) analysis provide superior sensitivity and detect changes at a greater resolution [6]. The aims of this study were to determine (1) continuous gait inter-limb coordination and stability levels of Veterans and SMs with TTA, and (2) the extent to which these levels are influenced by different Energy Storing and Returning (ESR) ankle-foot devices (i.e., ESR, articulating ESR (ART), and powered ESR (PWR)). It was hypothesized that individuals with vs. without TTA will indicate lower levels of coordination and stability, and that the PWR device will be associated with greater levels of coordination and stability compared to the ESR and ART device types.

CLINICAL SIGNIFICANCE

Improved coordination and stability due to a specific ankle-foot device will support evidence based guidelines for prosthetic prescription.

METHODS

Thirty individuals with unilateral TTA (55.3±13 years, 15 males, 1 female) were fit and evaluated with 3 different prosthetic ankle-foot devices: ESR, ART, and PWR. Participants were randomly assigned and separately utilized each prosthetic foot for 1 week at home. After each 1-week trial, participants with TTA and 10 age-matched individuals without TTA (control) participants underwent full-body biomechanical gait analysis. All participants walked at 1.3 m/s across a 10-meter instrumented walkway until at least 15 steps per foot were recorded. RP analysis calculated continuous measures of coordination, Mean Absolute Relative Phase (MARP), and stability, Deviation Phase (DP), between limbs (i.e., arms, legs, ipsilateral arms and legs, and contralateral arms

Table 1: Deficits in coordination and stability in Veterans and SMs with TTA compared to individuals without TTA.

	DP	Z	p-value
Arms			
Control	7.52(3.30)	-1.99	**0.04**
ART	8.55(6.47)		
Control	7.52(3.30)	-2.19	**0.03**
ESR	11.26(6.49)		
Legs			
Control	2.77(0.58)	-1.99	**0.04**
ART	3.64(0.68)		
Intact-Side Arm and Leg			
Control	6.09(2.52)	-2.09	**0.03**
ART	7.17(3.60)		
Intact-Side Arm and Prosthetic-Side Leg			
Control	6.48(3.62)	-2.45	**0.01**
ESR	8.80(4.08)		
Prosthetic-Side Arm and Intact-Side Leg			
Control	5.13(2.94)	-2.19	**0.03**
ESR	9.45(5.90)		
	MARP	**Z**	**p-value**
Arms			
Control	165.63(7.30)	-2.19	**0.03**
ESR	157.57(16.97)		
Intact-Side Arm and Leg			
Control	158.52(6.01)	-2.09	**0.03**
ESR	153.20(10.83)		

Note: Median(Interquartile Range). Significant values are shown in bold.

and legs). A low MARP value (closer to 0°) indicates a more in-phase relationship, while a high MARP value (closer to 180°) indicates a more anti-phase relationship. A low DP value (closer to 0°) indicates a more stable organization of the neuromuscular system and a high DP value (closer to 180°) indicates less stability. Due to non-normal distribution, Wilcoxon Signed Rank tests were used to determine significance at $p<0.05$.

RESULTS

Aim 1: Veterans and SMs with TTA experience deficits in coordination and stability compared to individuals without limb loss (Table 1). *Aim 2*: Veterans and SM with TTA had lower levels of coordination and stability with the ESR compared to the PWR and ART devices (Figure 1).

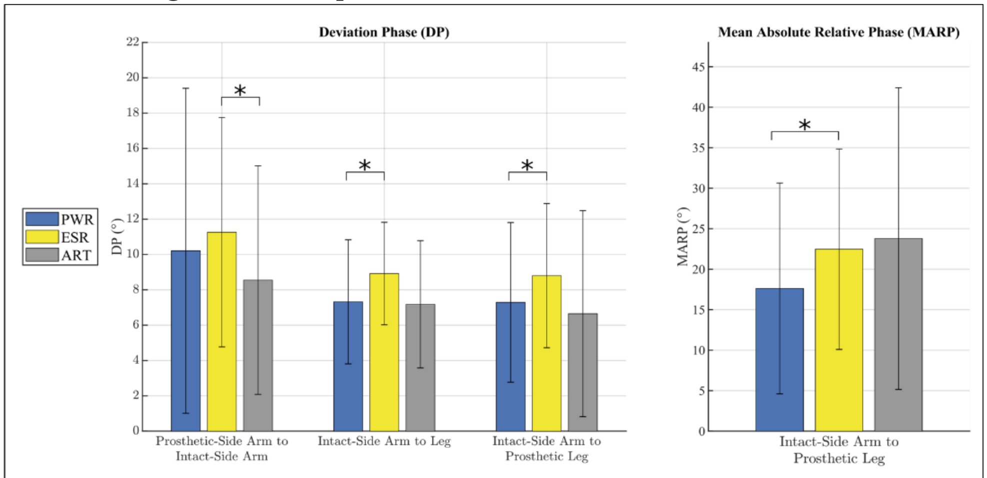

Figure 1: Significant differences in coordination and stability between devices.

DISCUSSION

Preliminary analysis of this dataset indicates that RP analysis is sensitive enough to identify differences between the experimental and control groups and device types. In contrast of our hypothesis, lower levels of coordination and stability observed with the ESR device were associated with the intact-side arm. This may highlight a compensatory effect of the intact-side arm swing during gait. However, further investigation is warranted, as continued analysis may indicate additional differences in coordination and stability between individuals with and without TTA and between device types. Importantly, due to the positive impact of coordination and stability on functional mobility and gait, findings from this study can directly influence prescription guidelines to optimize healthcare for all Veterans and SMs with TTA.

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DISCLOSURE STATEMENT

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EFFECTS OF A PASSIVE ANKLE FOOT ORTHOSIS ON ANKLE DORSIFLEXION IN PATIENTS WITH CLASSICAL FOOT DROP

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INTRODUCTION

Patients with classical foot drop (CFD), or drop foot syndrome, experience foot slapping after initial contact and foot dragging during the swing phase due to weak ankle dorsiflexion capacity [1]. To improve the biomechanical gait patterns and energy consumption, an ankle foot orthosis (AFO) is widely used as a clinical intervention to assist in ankle dorsiflexion [2]. The current problem associated with the usage of conventional AFOs is that normal walking gait patterns are disturbed due to the ankle's limited ankle movements [3], inhibition of a normal push-off during walking [4], and reduction of gait adaptability [5]. The goal of this study is to evaluate the effects of a passive AFO on ankle dorsiflexion in patients diagnosed with CFD.

CLINICAL SIGNIFICANCE

This study provides new evidence in support of a passive AFO as an assistive device to correct foot drop in patients with CFD.

METHODS

We recruited seven subjects (one male and six females, average age: 52 years) diagnosed with CFD for this study. All participants wore a commercially available passive AFO (XTERN, Turbomed, Orthotics, Quebec City, QC, Canada) on the affected limb(s) (Table 1). The participants were asked to walk over ground at their self-selected speeds with and without the usage of the passive AFO. 3-D marker trajectories, electromyography (EMG), and ground reaction force (GRF) data were collected during the walking trials. A 17-camera motion capture system (Vicon, Oxford, UK) sampled at 100 Hz was used. We placed sixty-one retro-reflective markers on the participants and measured joint kinematics through Vicon Nexus processing (CGM-2.5 PIG model).

Table 1: Demographic information of the participants

Participants	Gender	Clinical Diagnosis	Affected limb(s)
1	F	Peroneal nerve damage due to radiation	Left
2	F	Disc herniation in the L4/5 region	Left
3	M	Charcot-Marie-Tooth (CMT) disorder X1	Both
4	F	Nerve damage due to back surgery	Right
5	F	CMT X1	Both
6	F	Guillain-Barre syndrome	Both
7	F	No caused determined yet	Left

RESULTS

The average gait ratio of all affected limb (stance phase/gait cycle) have increased from 0.63 (SD: 0.042) to 0.65 (SD: 0.025) ($p = 0.03$) with the usage of the AFO. In addition, ankle joint kinematics showed an increase in dorsiflexion during different portions of the gait cycle for all affected limbs with the AFO (Figure 1). This is especially true, particularly within the swing phase (63-100% w/o AFO and 65-100% w/ AFO) of the gait cycle; in all subjects, the dorsiflexion increased substantially during the swing phase (Fig. 1). During the swing phase, the minimum average ankle angles increased from -29.6° (SD: 18.0°) to 1.5° (SD: 3.8°) ($p < 0.01$) and maximum average ankle angles improved from 1.7° (SD: 6.2°) to 10.9° (SD: 3.4°) ($p < 0.01$).

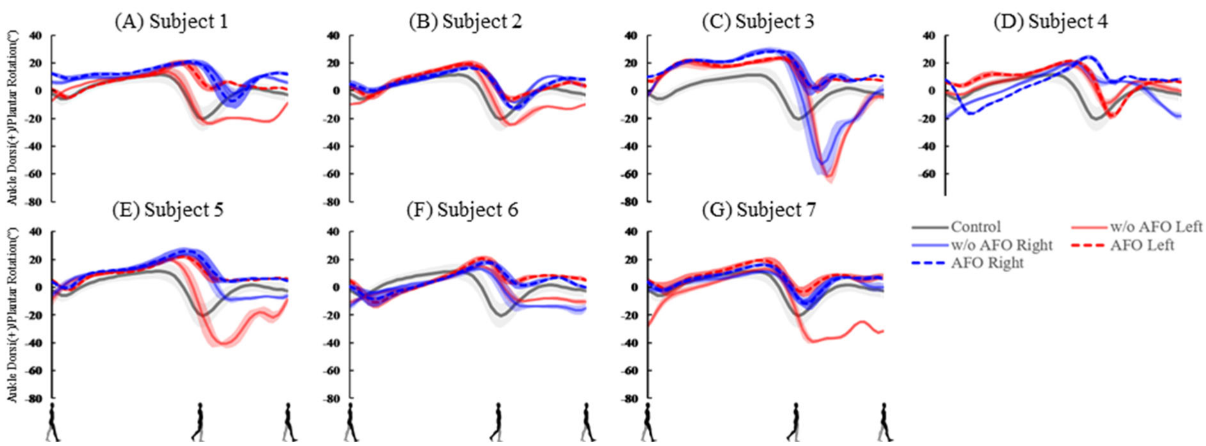

Figure 1. A passive AFO improved dorsiflexion for the affected limbs and normalized ankle kinematics to match more closely that of the able-bodied population. (±1 standard deviation is represented by the shaded bands).

DISCUSSION

Our results demonstrate that the passive AFO increased ankle dorsiflexion during gait when compared to walking without the device. These increases were most significant during the swing phase. These results suggest that this passive AFO aids with ankle dorsiflexion in patients diagnosed with CFD.

DISCLOSURE STATEMENT

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Effects of Powered Ankle-Foot Prosthesis on Intact Knee Adduction Moment

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INTRODUCTION

Individuals with a transtibial amputation (TTA) who use traditional energy storing and returning (ESR) devices often experience increased kinematic and kinetic asymmetries, particularly on their intact limb [1,2]. Research has correlated increased intact joint discomfort and knee osteoarthritis (OA) to abnormally high 1st peak knee adduction moments (KAM) of the intact limb [1,3]. Experienced TTA prosthesis users have been shown to have increased intact KAM of 46% and 17% compared to the prosthetic side and normative data, respectively [3]. Advanced prosthetic componentry, such as powered ankle-foot devices that mimic the gastroc-soleus complex, have the potential to reduce compensatory loading of the intact limb, which could reduce the risk of musculoskeletal injuries. The Empower foot (PWR) (Ottobock, Inc.) is the only commercially available prosthetic device able to provide the user with biomimetic power generation at push off. The aim of this study was to analyze the gait of individuals with TTA with different prosthetic foot conditions. We hypothesized that use of a PWR device would significantly reduce the 1st peak of the intact KAM with respect to an ESR device and would be normalized to the control group.

CLINICAL SIGNIFICANCE

Powered prosthetic devices could be used as a preventative measure to reduce compensatory affects during gait or to mitigate risks of knee OA caused by high 1st peak KAM within the intact limb of those with a TTA.

METHODS

Twenty-two individuals with TTA (age 52 ± 14.5 years, height 1.79 ± 0.1 m, and weight 87.6 ± 16.6 kg) with at least 1 year of prosthesis experience, and 8 control subjects were recruited from Veteran Affairs New York Harbor Healthcare System (NYHHS) and Walter Reed National Military Medical Center (WRNMMC). Study procedures were approved by each site's respective Institutional Review Board. Participants were randomized to wear both an ESR and PWR device, each for a 1-week acclimation period followed by a biomechanical gait evaluation. Control participants underwent a single biomechanical gait evaluation. Motion and force data were collected using an optical motion capture system (Qualisys, Goteborg, Sweden) and force platform system (AMTI, Waterford, MA). Participants were fit with 78 retroreflective markers placed symmetrically about the entire body. Participants performed level ground walking at 3 speeds: 1.0, 1.3, and 1.5 m/s. Data was analyzed using Visual3D software (C-Motion Inc.) to derive kinematic, kinetic, power, and ground reaction force results. A one-way, between subjects, repeated measures analysis of variance (ANOVA) was used to determine if significant differences (p<0.05) were present for the 1st peak intact KAM between each foot and between individuals with and without TTA for all speeds.

RESULTS

The average time since amputation was 15.6 ± 13.7 years. The major cause of amputation was trauma (72%), followed by cancer (17%), and dysvascular disease/diabetes (11%). The control participants consisted of all male individuals, age 32 ± 4.8 years, height 1.78 ± 0.05m, and weight 85.7 ± 12.4 kg. First peak intact KAMs for the PWR device at each speed (1.0, 1.3, and 1.5m/s) were not significantly lower than the ESR device (p>0.05). Table 1 shows the average 1st peak intact KAMs (Nm/Kg) for ESR (0.45, 0.52, 0.59), PWR (0.45, 0.52, 0.58), and the Controls (0.46, 0.47, 0.54) at 1.0, 1.3 and 1.5m/s, respectively. The one-way between subject, repeated measures ANOVA showed each were not significantly different (p>0.5). Upon further analysis, there were no significant difference in KAM between the prosthetic devices and the control participants at any speed (p > 0.05).

1st Peak Knee Adduction Moment (Nm/kg) and % Differences Between Feet and Control						
Moment	N = 22		N = 8	Difference		
Speeds (m/s)	ESR	PWR	Control	Control to ESR	Control to PWR	PWR to ESR
1.0	0.45	0.45	0.46	1.0	1.0	0
1.3	0.52	0.52	0.47	-5.0	-5.0	0
1.5	0.59	0.58	0.54	-5.0	-4.0	-1.0

Table 1: 1st Peak KAM and % Differences Between Feet and Control.

DISCUSSION

Biomimetic devices like the Empower can generate normative ankle power during push off and have been previously reported to reduce peak KAM of the intact knee [1]. Previous research has shown that the 1st peak of the intact KAM for individuals with TTA who utilized a powered device was significantly reduced by 20.6% and 12.2% at walking speeds of 1.5 and 1.75 m/s (p = 0.03, 0.05) compared to the ESR condition [1]. However, at slower walking speeds, comparable to the speeds used in this investigation, KAM was not significantly different between the ESR and PWR devices. These differences may be attributable to the net positive work performed at faster speeds by the PWR, whereas at slower speeds the net mechanical work is nearly zero across the entire stance phase [1]. Furthermore, while the PWR device provided biomimetic push-off power, the uniaxial movement cannot replicate the biarticular nature of the gastrocnemius, which may reduce the efficiency of the load transmission to the intact limb [2]. As such, the reduction in intact KAM reported in this investigation could be the result of continued compensatory strategies of the intact limb and reduced stability of the knee in the frontal plane. Additionally, the average age of the control subjects was approximately 20 years younger than the participants with TTA, which may influence the 1st peak KAM for control subjects. Future work will include age-matched control subjects along with additional TTA participant KAM data.

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DISCLOSURE STATEMENT

The views and opinions expressed herein do not necessarily state or reflect those of the Departments of Defense or Veterans Affairs and shall not be used for advertising or product endorsement purposes. The authors have no conflicts of interest.

Upper Extremity Motion Analysis After Scapular Arthrodesis

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PATIENT HISTORY

The patient is a 13-year-old female with fascioscapulohumeral muscular dystrophy (FSHD) with bilateral upper extremity involvement. The patient had difficulty raising her arms and activities of daily living (ADL) such as brushing her hair was challenging. The surgical course started with a staged scapular arthrodesis on the dominant right scapula, followed by surgical correction of the left scapula 12 months later.

CLINICAL and MOTION DATA

Physical Exam, Pediatrics Outcomes Data Collection Instrument (PODCI) and three-dimensional motion analysis (3DMA) data were collected preoperatively and 1 month following the 2nd surgical procedure. The physical exam included 2D goniometry of active and passive shoulder motion. The 3DMA measured the motion of the patient's shoulder while she performed the functional tasks of active shoulder abduction, raising a ball, moving the hand to touch the small of the back, and to touch the neck.

TREATMENT DECISIONS AND INDICATIONS

The treatment for patients with FSHD who have incapacitating shoulder weakness due to abnormal scapula motion is scapulothoracic arthrodesis.

OUTCOME

Postoperatively physical examination demonstrated improvements in active and passive shoulder motion (Table 1). PODCI Upper Extremity subtest improved by 17, Pain and Comfort improved by 20, and Global Functioning improved by 8 points. Motion analysis showed a 22^{o} improvement in right shoulder abduction, a 12^{o} improvement in left shoulder abduction, a 22.3^{o} improvement in right shoulder flexion and a 4.4^{o} improvement in left shoulder flexion. (Figure 1) Bilateral shoulder extension is now normal during hand to back activities. The patient is now able to perform upper extremity dressing independently. She continues to require some assistance for brushing her hair.

SUMMARY

FSHD is a rare disease characterized by progressive weakness of the serratus anterior, trapezius, and rhomboid muscles, sparing the deltoid and rotator cuff. Scapulothoracic arthrodesis stabilizes the scapula for improved movement. Upper extremity motion analysis does provide objective data on upper extremity function in multiple planes and is useful in patients with FSHD to understand the progression of their disease course and function before and after arthrodesis.

DISCLOSURE STATEMENT

The authors have no conflicts of interest to disclose.

Table 1: Preoperative and Postoperative Physical Exam Results

	Range of Motion	Pre - operative	Post - operative	difference
Right Shoulder Abduction	Active	60°	90°	30.0°
	Passive	110°	145°	35.0°
Left Shoulder Abduction	Active	60°	60°	0.0°
	Passive	110°	145°	35.0°
Right Shoulder Forward Flexion	Active	40°	70°	30.0°
	Passive	95°	120°	25.0°
Left Shoulder Forward Flexion	Active	40°	60°	20.0°
	Passive	95°	120°	25.0°

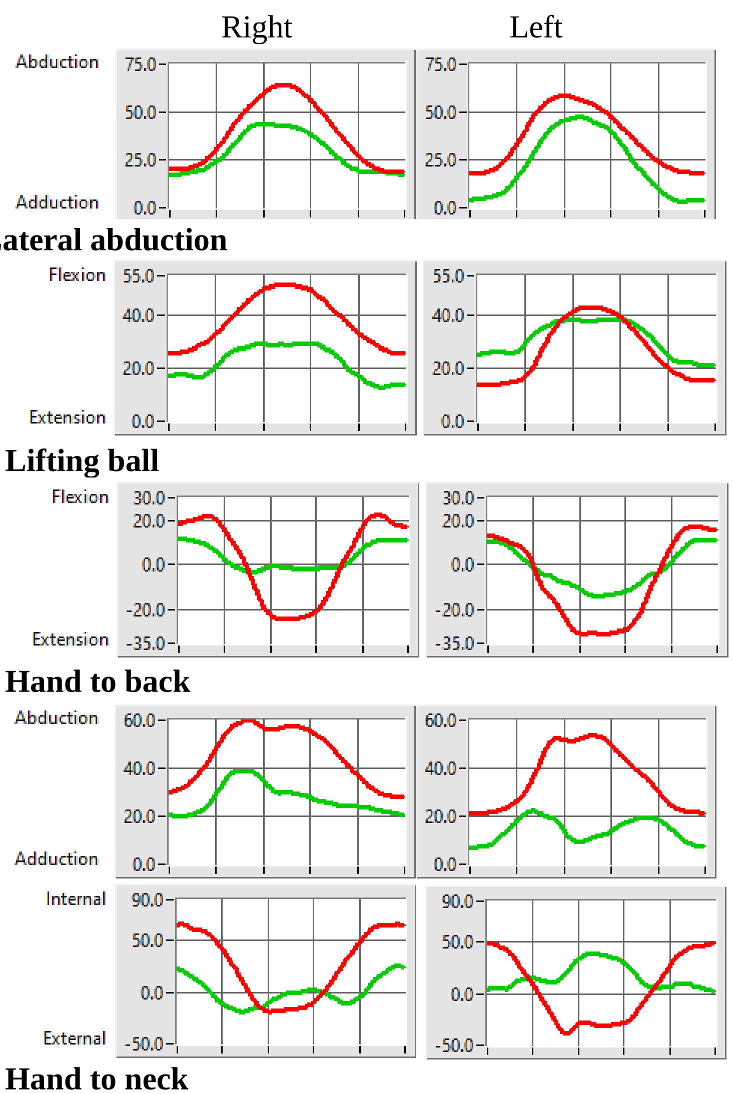

Figure 1: Shoulder Kinematics Green = pre-op, Red = post-op.

SHOULDER MOTION OVERESTIMATED FROM CLINICAL MALLET SCORES

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INTRODUCTION

Brachial plexus birth injuries (BPBI) occur in up 0.9 to 4.6 per live 1,000 births[1]. Weakness about the shoulder and development of glenohumeral joint contractures are common sequalae of BPBI[2]. These contractures result from strength imbalances between the internal rotators and the external rotators[2], as well as reduced muscle growth[3]. Shoulder function in children with BPBI is frequently assessed using the modified Mallet classification (Figure 1) to evaluate upper extremity motion deficits[4]. Performance of each position is classified by the clinician. However, the modified Mallet classification has been criticized for lack of specificity and large variation within single scores in the ordinal scale[4,5,6].

The purpose of this study was to assess the accuracy of the abduction and external rotation Mallet classification scores in children with BPBI. These positions were used because the scores are defined by discrete joint angles. We hypothesized that Mallet scores determined by clinical exam would differ from Mallet scores determined by motion capture joint angle measurements.

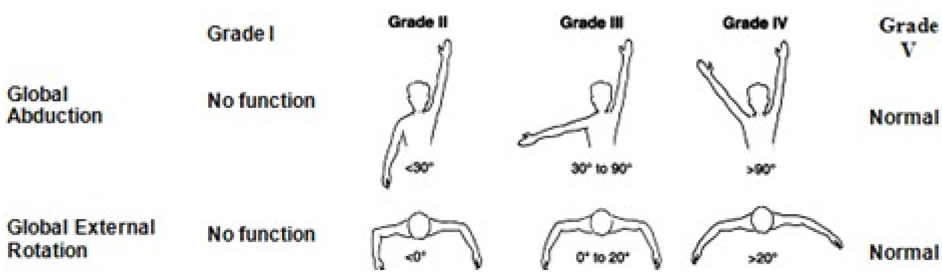

Figure 1: Abduction and external rotation portions of the Mallet classification

CLINICAL SIGNIFICANCE

Mallet scores are commonly used for clinical decision making and are routinely reported in the literature for intervention outcomes assessment. Consequently, accurate determination of Mallet scores is critical for clinical care of children with BPBI.

METHODS

107 children with BPBI who had a motion capture assessment and Mallet scores determined for clinical purposes on the same date were reviewed retrospectively. The Mallet scores that assess planar measurements with defined joint angles were included (abduction and external rotation). The clinical Mallet scores were obtained from the medical record. Mallet score assessments were performed by a hand surgeon or occupational therapist specialized in pediatric upper extremity care with substantial experience with Mallet scores. Motion capture measurements were used to calculate humerothoracic elevation and external rotation joint angles in the abduction and external rotation positions, respectively. The

humerothoracic joint angles were converted to the corresponding Mallet scores. The Mallet scores determined by the motion capture-measured joint angles were considered the gold standard. Discrepancies between the Mallet scores determined by clinicians and those determined by motion capture were assessed.

RESULTS

For abduction, 24.3% of Mallet scores were misclassified during clinical examination. Of the misclassified scores, 22 were overestimated by 1 point and 4 were underestimated by 1 point compared to motion capture. For external rotation, 72.9% of Mallet scores were misclassified during clinical examination. Of the misclassified scores, 43 were overestimated by 1 point, 34 were overestimated by 2 points, and 1 was underestimated by 1 point.

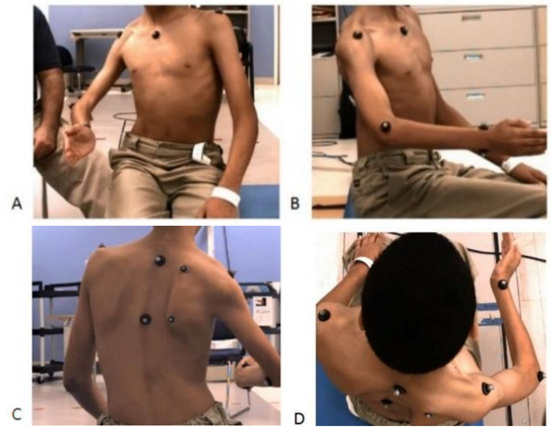

Figure 2: In the external rotation position viewed from the front (A) and side (B), the patient appears to have humerothoracic rotation near neutral. However, the posterior view (C) emphasizes the lateral flexion of the thorax. The top view (D), helps to demonstrate that the humerus is actually in internal rotation relative to the thorax.

DISCUSSION

Children with BPBI were observed to utilize many compensatory movement strategies to maximize their range of motion. These include altered scapulothoracic movement patterns as well as thoracic movements, such as leaning, to accomplish greater reach. Children with BPBI also frequently elevate their arms and extend their wrists giving the false appearance of greater external rotation (Figure 2). These compensatory strategies may be difficult to account for during clinical exams, which can ultimately lead to overestimation of Mallet scores. Prior studies have demonstrated acceptable intra- and inter-rater reliability of the Mallet classification[7,8] however, reliability does not imply accuracy. The inaccuracy of the clinically determined Mallet scores is alarming given that they are frequently utilized to assist with surgical indications and are one of the most common outcome measures utilized in the BPBI literature. Objective measures, such as motion capture, offer an alternative for accurate assessment of shoulder function in children with BPBI.

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DIFFERENCE IN BILATERAL UPPER LIMB MUSCLE ACTIVATION DURING ASSISSTING HAND ASSESSEMENT IN CHILDREN AFFECTED BY UNILATERAL CEREBRAL PALSY

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INTRODUCTION

Cerebral palsy (CP) refers to a group of childhood-onset disorders that affect an individual's ability to move and maintain posture due to a brain lesion occurring during in utero development. When the lesion affects motor areas on one side of the brain, an impairment of movement and posture will be manifested on the contralateral side. The most common CP condition is called unilateral cerebral palsy (UCP), leading to challenges with daily activities due to decreased function [1] and use of their affected upper limb. The purpose of this study was to evaluate side-to-side muscle activation differences during the Assisting Hand Assessment (AHA), a performance based assessment that measures how children with upper limb impairments use their involved limb as an assisting hand during bimanual activities [2].

CLINICAL SIGNIFICANCE

As a visual tool, the AHA provides qualitative information regarding the upper extremity functionality of a child with UCP. Discrepancies in side-to-side muscle activation provides a way to quantify the AHA tasks that is unattainable with visual analysis alone.

METHODS

Thirteen patients diagnosed with UCP were tested. Muscle activation was measured via electromyography (EMG) sensors (Delsys Trigno Wireless, Boston, MA, USA) placed bilaterally on the biceps (BIC), triceps (TRI), wrist extensors (EXT), and wrist flexor (FLE) muscles. Tasks of the AHA were assessed individually as shown in *Table 1*. The EMG sensors measured the electrical activity of each muscle in micro-volts (μV) at 3,000 Hz. EMG signals were rectified and filtered to compute a linear envelope, and mean and maximum values were computed for both limbs using a representative trial per task. To determine side-to-side differences, a symmetry ratio (affected vs. unaffected) was computed and Wilcoxon signed-ranked tests were performed to identify significant differences using an α level of 0.05.

Table 1: AHA Tasks

AHA Task	Primary Movement
Car	Forearm reach
Marbles	Lateral reach distance
Brewing	Wrist activation
Antenna	Shoulder flexion
Rip Paper	Forearm rotation
Medal	Shoulder flexion
Color & Cut	Elbow flexion
Music	Elbow flexion
Cymbals	Shoulder flexion
Large Cymbals	Shoulder flexion
Bracelets	Wrist rot., elbow flex.
String toy	Lateral reach distance

RESULTS

Side-to-side differences in muscle activation were observed across tasks, primarily for the biceps, wrist extensor, and wrist flexor muscle groups. Specifically, differences in biceps and wrist flexor muscle activation were observed in 50% and 42% of tasks, respectively, while wrist extensor muscle activation significantly differed between sides across all tasks. Notably, only three tasks resulted in side-to-side differences solely in the wrist extensor muscle group, including the *Car* (p = 0.001), *Cymbals* (p = 0.016), and *String Toy* (p = 0.004) tasks.

Table 2: Side-to-Side muscle activation

Tasks	BIC	TRI	EXT	FLE
Car	0.8 ± 0.5	1.7 ± 1.2	**0.5** ± **0.2**	0.9 ± 0.7
Marbles	0.9 ± 0.7	1.6 ±1.2	**0.4** ± **0.2**	**0.6** ± **0.4**
Brewing	**0.9** ± **1.1**	1.2 ± 1.3	**0.6** ± **0.3**	1.1 ± 0.8
Antenna	1.8 ± 4.4	1.4 ± 2.3	**0.7** ± **0.4**	**0.8** ± **0.8**
Rip Paper	**0.8** ± **0.8**	**0.8** ± **0.8**	**0.6** ± **0.3**	**0.9** ± **1.4**
Medal	0.9 ± 0.9	0.8 ± 0.7	**0.5** ± **0.1**	**0.5** ± **0.4**
Color & Cut	**1.0** ± **1.3**	0.9 ± 0.5	**0.6** ± **0.3**	0.8 ± 0.5
Music	**0.7** ± **0.6**	1.2 ± 0.9	**0.3** ± **0.6**	**0.7** ± **0.5**
Cymbals	0.9 ± 0.8	1.2 ± 0.6	**0.4** ± **0.8**	0.6 ± 1.0
Large Cymbals	**0.7** ± **0.6**	1.0 ± 0.6	**0.8** ± **0.8**	1.0 ± 0.6
Bracelets	**0.7** ± **0.6**	1.6 ± 1.9	**0.6** ± **0.3**	0.8 ± 0.4
String Toy	0.7 ± 0.6	1.2 ± 0.9	**0.5** ± **0.2**	0.8 ± 0.5

Note: Shaded cells indicate less activation on the affected side, and bold text indicates significant side-to-side differences.

Alternatively, triceps differences were not significant with the exception of the *Rip Paper* task (p = 0.019). Symmetry ratios for the triceps also indicated increased activation on the affected side for 9 of the 12 tasks, while the affected side exhibited decreased activation compared to the unaffected side for the majority of tasks in the remaining muscle groups.

DISCUSSION

These results indicate that side-to-side differences in muscle activation are more exaggerated in specific tasks of the AHA, especially in the *Rip Paper* task that involves both limbs to move similarly in opposite directions. Therefore, certain tasks may provide more information to guide rehabilitation that focuses on muscle recruitment and strength. Additionally, while wrist extensor muscle activation differed bilaterally across all tasks, side-to-side differences in the remaining muscle groups varied across tasks, indicating the importance of evaluating the full AHA rather than focusing on one or two components of the assessment. It is also important to note that differences in triceps muscle activation could have been limited due to a lack of AHA tasks requiring substantial elbow extension and thus, deep triceps muscle contraction. Future work will explore the use of motion capture to quantify additional compensatory mechanisms such as trunk displacement.

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DISCLOSURE STATEMENT

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ASSESSMENT OF UPPER EXTREMITY REACHABLE WORKSPACE IN CHILDREN WITH BRACHIAL PLEXUS BIRTH INJURY

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INTRODUCTION

Approximately 1 in every 1,000 live births results in lifelong impairments in upper extremity (UE) function from brachial plexus birth injury (BPBI) with deficits in shoulder elevation, external rotation, and extension being most common [1]. Clinical interventions are often employed to improve UE function. Accurate quantification of the entirety of an individual's UE function is critical for informing patient-specific decision-making and outcomes assessment. Current clinical assessments attempt to capture an individual's UE functionality using a limited set of static postures and/or movements and have been criticized for being insensitive to certain meaningful differences in function [2].

Reachable workspace provides a numeric and visual assessment of global UE function by quantifying the regions in space that patients can reach with their hands [3]. Incorporation of real-time visual feedback into workspace collection can engage patients to provide a 'best effort' acquisition of UE function in regions surrounding their body [4]. Reachable workspace can differentiate between patients and healthy controls [5], but has not been assessed in BPBI.

This study utilizes a real-time feedback reachable workspace tool to quantify outer, far-from-body UE function on children with unilateral BPBI. It was hypothesized that the affected limb would have less workspace than the unaffected limb and that differences would be greatest in regions requiring shoulder elevation and/or external rotation, and extension.

CLINICAL SIGNIFICANCE

Reachable workspace may provide a more complete measure of global UE function while giving an intuitive visual depiction of patient function for clinicians, caretakers, and patients.

METHODS

Trunk and UE segment orientations of 21 children with unilateral BPBI (3-12 years; various injury levels and surgical histories) were measured with motion capture (VICON, Centennial, CO). An array of virtual targets surrounding the participant was created using spherical coordinates. Custom software displayed targets and real-time visual feedback from motion capture (**Fig. 1A**). Movements of a red cursor sphere were controlled in real-time based on the position of the participant's hand marker relative to the

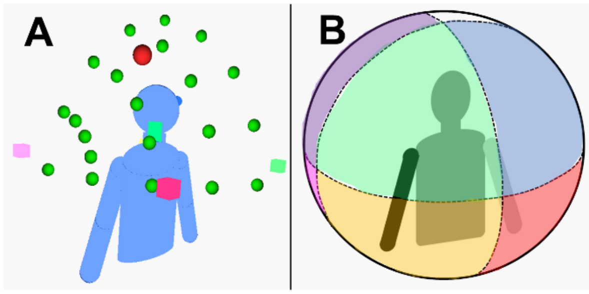

Figure 1: A) real-time feedback displayed. B) Six octants analyzed – anterior (blue/green/red/yellow), superior (blue/green/purple), ipsilateral (green/yellow/purple/pink).

trunk (**Fig. 1A**). Targets throughout the spherical space were displayed sequentially by octant region (e.g., superior hemisphere, right side, anterior) (**Fig. 1B**) until all targets were completed or until the participant was unable to reach any more; superior/inferior posterior contralateral octants were not analyzed due to their limited workspace. Participants completed trials on their affected and unaffected limbs. Percent workspace reached was calculated for each octant region as the points reached by the hand expressed as a percentage of potential reachable points along the outer surface area. Median reach distance was also calculated as a percent of arm length for each octant. Percent workspace reached and median reach distance were each evaluated with a 3-way repeated measures ANOVA with post hoc analyses.

RESULTS

The affected limb had significantly less percent workspace reached and median reach distance than the unaffected limb for all six octant regions (**Fig. 2**).

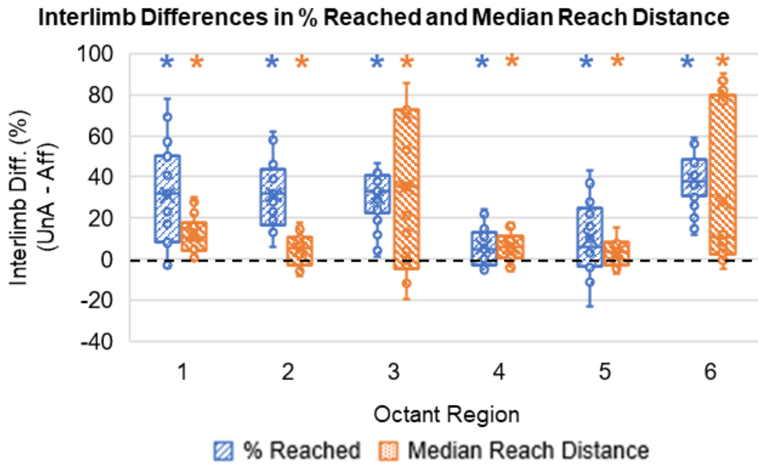

Figure 2: Interlimb differences (unaffected – affected) in % workspace reached and median reach distance by octant region (1=sup./ant./ipsil., 2= sup./ant./contra., 3=sup./post./ipsil., 4=inf./ant./ipsil., 5=inf./ant./contra., 6=inf./post./ipsil. Positive values indicate greater workspace on the unaffected limb. * indicates significance (p<0.05).

DISCUSSION

Interlimb differences in percent workspace reached were largest in octants requiring shoulder elevation (upper anterior ipsilateral and contralateral), a combination of elevation and external rotation (upper posterior ipsilateral), and extension (lower posterior ipsilateral). Affected side median reach distance was also decreased throughout, consistent with the high frequency of elbow flexion contractures in this population. These findings suggest that reachable workspace is capable of characterizing the nature of the UE impairments commonly found on the affected limb in BPBI and demonstrate the capacity of a real-time feedback reachable workspace tool to assess UE function in BPBI.

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DISCLOSURE STATEMENT

The authors have no conflicts of interest to disclose.

EVALUATION OF UPPER EXTREMITY REACHABLE WORKSPACE IN CHILDREN WITH SPASTIC, HEMIPLEGIC CEREBRAL PALSY

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INTRODUCTION

Cerebral palsy (CP) occurs in about 1.5-4 of every 1,000 live births with roughly one-third of patients having spastic, hemiplegic CP [1]. Unilateral upper extremity (UE) functional impairments due to limitations in arm movement stem from spasticity alone or spasticity with concurrent joint contractures [2]. Regarding the shoulder and elbow, shoulder abduction and/or external rotation and elbow extension are most frequently limited [2]. A variety of interventions are employed to improve UE function. Accurate quantification of the entirety of an individual's UE function is critical for informing patient-specific decision-making and outcomes assessment. Current clinical assessments attempt to capture UE function using a limited set of static postures and/or movements and have been criticized for being insensitive to certain meaningful differences in function [3]. Reachable workspace provides a numeric and visual assessment of global UE function by quantifying the regions in space that patients can reach with their hands [4]. Workspace can distinguish between patients and healthy controls [5], but has not been assessed in CP.

A real-time feedback reachable workspace tool [6] was used to quantify outer, far-from-body UE function on children with spastic, hemiplegic CP. It was hypothesized that the affected limb would have less workspace than the unaffected limb and that differences would be greatest in areas requiring shoulder abduction and/or external rotation and elbow extension.

CLINICAL SIGNIFICANCE

Reachable workspace may provide a more complete measure of global UE function while giving an intuitive visual depiction of patient function for clinicians, caretakers, and patients.

METHODS

Trunk and UE segment orientations of 17 children with spastic, hemiplegic CP and minimal to no cognitive impairment (4-18 years) were measured with motion capture. An array of virtual targets surrounding the participant was created using spherical coordinates. Custom software displayed targets and real-time visual feedback from motion capture (**Fig. 1A**). The red sphere's movement was controlled in real-time by the position of the participant's hand marker relative to the trunk (**Fig. 1A**). Targets throughout the spherical space were displayed sequentially by

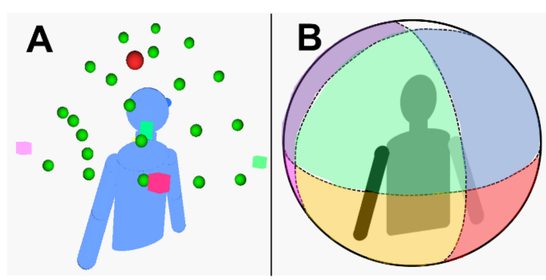

Figure 1: A) real-time feedback displayed. B) Six octants analyzed – anterior (blue/green/red/yellow), superior (blue/green/purple), ipsilateral (green/yellow/purple/pink).

octant region (e.g., superior hemisphere, right side, anterior) (**Fig. 1B**) until all targets were completed or until the participant was unable to reach any more; superior/inferior posterior contralateral octants were not analyzed. Participants completed trials on their affected and unaffected limbs. Percent workspace reached was calculated for each octant region as the points reached by the hand as a percentage of potential reachable points along the outer surface area. Median reach distance was calculated as a percentage of arm length. Each measure was evaluated with a 3-way repeated measures ANOVA with post hoc analyses.

RESULTS

The affected limb had significantly less percent workspace reached than the unaffected limb on the superior anterior contralateral, superior posterior ipsilateral, and inferior posterior ipsilateral octants (**Fig. 2**). The affected limb had significantly less median reach distance than the unaffected limb on the superior ipsilateral anterior and posterior octants (**Fig. 2**).

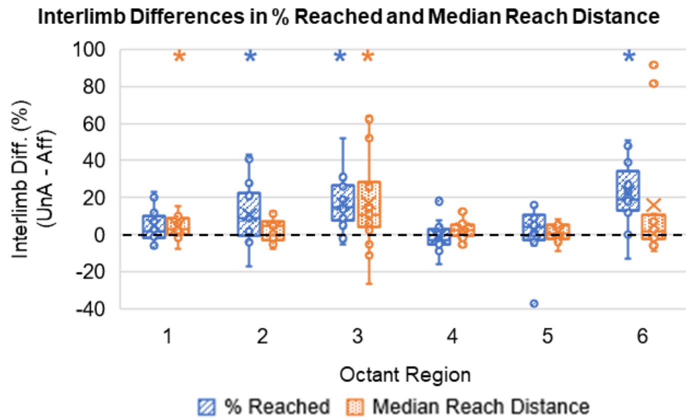

Figure 2: Interlimb differences (unaffected – affected) in % workspace reached and median reach distance by octant (1=sup./ant./ipsil., 2=sup./ant./contra., 3=sup./post./ipsil., 4=inf./ant./ipsil., 5=inf./ant./contra., 6=inf./post./ipsil. Positive values indicate greater workspace on the unaffected limb. * indicates significance ($p<0.05$).

DISCUSSION

This real-time feedback reachable workspace tool detected interlimb differences in reach in this mildly affected cohort of children with spastic, hemiplegic CP. The measured reach deficits were consistent with clinically observed differences in the affected arm. This cohort demonstrated the greatest difficulty with reaching behind (octants 3/6) and up and across (octant 2). These regions may represent more challenging motor patterns for children with CP as they require midline integration or reach outside the visual field. While further evaluation is needed in patients with more severe involvement, these findings suggest that reachable workspace can be a clinically relevant tool for evaluation of UE function in children with CP.

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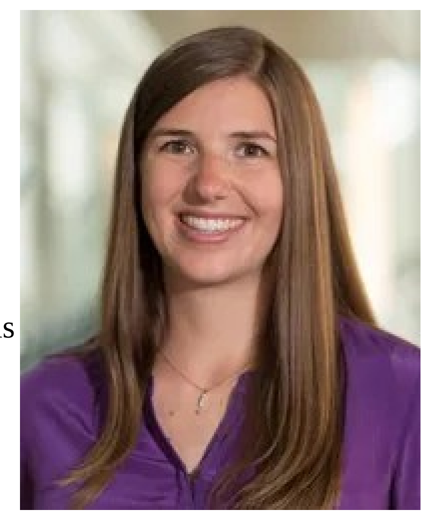
DISCLOSURE STATEMENT

The authors have no conflicts of interest to disclose.

Untangling control: Understanding the role of altered motor control on gait in cerebral palsy

Dr. Kat M. Steele

Cerebral palsy is caused by an injury to the brain near the time of birth. The fact that this injury impacts control is obvious. However, given the complex diversity of these injuries and the experiences of early childhood that shape motor pathways, understanding how control is altered for a given child and how these changes impact movement is an incredibly difficult challenge to untangle. In the upper-extremity new stimulation and imaging techniques are helping to characterize motor pathways and their impact on movement after injury. In the lower-extremity, the motor cortex is packed tightly within the longitudinal fissure, making it challenging to access with imaging or even just separate the left and right sides of the body. Further, spinal networks play a much larger role in control of the lower-extremity than the upper-extremity. These challenges pose unique barriers in our capacity to characterize or influence altered control. In this talk, I will untangle and attempt to piece together our current tools and understanding of altered control of gait in cerebral palsy. We'll examine how current tools in gait analysis, causal modeling, and stimulation can help untangle altered control. We'll highlight open gaps that need to be addressed to help us not just understand altered control, but leverage this knowledge to support movement and participation. There is still a tangled mess at the end, so we hope you will join with ideas, discussion, and debate.



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GCMAS2022 - Wednesday, June 8, 2022 – 1.00pm-2.00pm

EVALUATING FAMILY PRIORITIES BY AGE USING THE GAIT OUTCOMES ASSESSMENT LIST (GOAL) QUESTIONNAIRE

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INTRODUCTION: The Gait Outcomes Assessment List (GOAL) questionnaire is a valid, reliable outcome assessment of family priorities and functional mobility for ambulatory children with cerebral palsy (CP).[1-3] It distinguishes priorities by Gross Motor Function Classification (GMFCS) level.[3-4] Family priorities for mobility and intervention may also differ by age.[5] The purpose of this study was to determine if the GOAL questionnaire distinguishes family priorities by age within GMFCS levels.

CLINICAL SIGNIFICANCE: Understanding priorities by age within a functional level will assist clinicians in providing optimal family-centered care and may improve outcomes when incorporated into treatment planning.

METHODS: Caregivers of consecutive unique individuals with a diagnosis of CP (GMFCS I-III) completed parent version 5.0 of the GOAL questionnaire as part of a 3-D gait analysis in a tertiary care facility for children with disabilities. The proportion of respondents who indicated items to be at least 'difficult' to perform and 'very important' to improve were considered priorities and were summarized by age group within each GMFCS level. Chi-square analysis was used to test for differences with post-hoc comparisons.

RESULTS: Data from 471 individuals (284 M, mean age 10.1±3.7 years, range 4-17 years) were included. Age category and GMFCS sample sizes: 4-7 years:164; 8-12 years:139; 13-17 years:168; GMFCS I:131; II: 206; III: 134. Top priorities differed by age within each GMFCS category. The 4-7 year-old group was most frequently distinguished age category from other categories (Table). Priorities often shifted with age. For example, in the Gait Pattern and Appearance Domain priorities shifted from walking with feet flat (4-7y) to walking taller (8-12y, 13-17y). In the Gait Function & Mobility domain, shift occurred from walking faster (4-7y) to walking farther (8-12y, 13-17y) to walking for longer duration (13-17y). Four of 5 top priority items for 4-7 year-olds were significantly different in GMFCS I and 1 of 5 top priorities were significantly different in GMFCS level III. Limited statistical differences were present in GMFCS level II in the top 14 priorities. See Table.

DISCUSSION: The GOAL questionnaire identifies the differences in top priority goals based on age. When statistical significance is reached, the priorities for 4-7 year-olds were distinguished from the other age groups. Trends in additional items were noted that did not reach statistical significance. A larger dataset will solidify these findings. Use of the GOAL questionnaire to identify family priorities and how they change with age will be an important adjunct to shared decision-making as part of family-centered care.

Table:

MOST COMMON PRIORITIES BY AGE WITHIN GMFCS LEVEL (BY PROPORTION)																				
GMFCS I							GMFCS II							GMFCS III						
Item	all	4 to 7	8 to 12	13 to 17	chi sq	sig pair <.05	Item	all	4 to 7	8 to 12	13 to 17	chi sq	sig pair <.05	Item	all	4 to 7	8 to 12	13 to 17	chi sq	sig pair <.05
G47-move vs others	**0.33**	**0.42**	**0.39**	**0.25**			**C25 tired walking**	**0.54**	**0.57**	**0.60**	**0.46**			**E38 - walk not drag feet**	**0.53**	0.48	**0.53**	**0.59**		
C25 - tired walking	**0.29**	0.38	**0.36**	**0.18**			**E39 walk not trip/fall**	**0.43**	0.44	**0.47**	0.39			**C25 tired walking**	**0.51**	**0.60**	**0.48**	**0.41**		
E37 - walk feet straight	**0.27**	0.39	**0.30**	0.17			**B14 walk fast to keep up**	**0.43**	**0.46**	0.32	**0.52**			**B16 up/down stairs**	**0.49**	**0.62**	0.39	0.38		
F41 - braces/orthotics	**0.27**	0.40	**0.31**	0.13			**E37 - walk feet straight**	**0.43**	**0.46**	0.38	**0.46**			**E36 walk taller**	**0.49**	0.48	0.37	**0.61**		
E38 - walk not drag feet	**0.26**	**0.42**	0.27	0.15	X^2=7	4-7 vs. 13-17	**E35-walk feet flat**	**0.43**	**0.48**	0.38	0.41			**A9 in/out car**	**0.48**	**0.59**	**0.43**	0.36		
C26- tired other activities	0.25	**0.39**	**0.31**	0.12	X^2=9	4-7 vs. 13-17	G47-move vs others	0.42	0.39	**0.47**	0.40			B19 walk slippery	0.46	**0.55**	0.37	**0.41**		
E35-walk feet flat	0.25	**0.41**	0.18	0.18	X^2=7	4-7 vs. 13-17	B19 walk slippery	0.42	**0.45**	0.38	0.42			B10 walk >250 meters	0.46	0.48	**0.47**	**0.41**		
D29-ride bike/trike	0.24	**0.41**	0.26	0.11	X^2=8	4-7 vs. 13-17	E36 walk taller	0.41	0.28	**0.45**	**0.51**	X^2=7	4-7 vs. 13-17	E35-walk feet flat	0.43	0.50	0.37	0.37		
D33- balance activities	0.23	0.23	0.22	**0.25**			E38 - walk not drag feet	0.40	0.32	0.39	**0.49**			B15 avoid obstacles	0.40	0.53	0.24	0.34	X^2=7	4-7 vs. 8-12
E36 walk taller	0.22	0.24	0.16	**0.25**			D33- balance activities	0.38	**0.48**	**0.42**	0.25			E37 - walk feet straight	0.40	0.45	0.33	0.39		
C20 pain feet/ankles	0.21	0.24	0.19	**0.20**			D27 run	0.37	0.30	0.41	0.40			A6 walk carry object	0.40	0.53	0.23	0.36	X^2=7	4-7 vs. 8-12
G48 others feel move	0.21	0.23	0.16	**0.22**			B10 walk >250 meters	0.36	0.43	0.38	0.27			B13 walk > 15 min	0.40	0.42	0.30	**0.46**		
D27 run	0.19	0.33	0.13	0.14	X^2=8	4-7 vs. 8-12; 4-7 vs.13-17	B13 walk > 15 min	0.34	0.42	0.32	0.28			A5 dress	0.40	**0.57**	0.31	0.21	X^2=12	4-7 vs. 8-12; 4-7 vs 13-17
D32 jumping sports	0.19	0.26	0.17	0.17			C26- tired other activities	0.34	0.28	0.39	0.35			G47-move vs others	0.39	0.37	**0.43**	0.40		

Proportion of respondents indicating an item was a 'very important' goal to improve and at least 'difficult' to perform.
Bolded proportions represent the top five priorities across all age levels and within a specific GMFCS level.
Top 14 of 49 priority items are displayed to capture all top priority items for each age group within a given GMFCS level.
Additional significant pairwise comparisons may be present in priorities 15-49 not presented here.
critical X^2 value=6.0

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DISCLOSURE STATEMENT

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PODCI Scores within GMFCS Level in Individuals with Cerebral Palsy

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INTRODUCTION: The Pediatric Outcomes Data Collection Instrument (PODCI) is a patient/parent reported outcome measure widely used in children with cerebral palsy (CP). PODCI scores have been compared for patients of different Gross Motor Function Classification System (GMFCS) levels [1]. However, PODCI score variability has not been examined in patients of GMFCS level IV. The purpose of this study is to examine the distribution of PODCI scores within patients with CP GMFCS level I-IV.

CLINICAL SIGNIFICANCE: PODCI scores have not been examined in this manner for individuals that are GMFCS level IV. Utilizing a patient/parent reported outcome measure is useful for providing patient centered care for areas that are meaningful to the family.

METHODS: We retrospectively identified all patients with CP seen for gait analysis at our institution who had been assigned a GMFCS level and whose parent/caregiver had completed the PODCI at their visit. A total of 214 subjects met the eligibility requirements, 35 were GMFCS I, 105 were II, 43 were III, and 31 were IV. Mean age was 11.1, SD 3.7 years (range 4.1-18.9). When patients had more than one visit, their first visit with available data was selected. Patients were grouped based on GMFCS level. One-way ANOVA with pairwise Bonferroni-adjusted post hoc tests was performed to compare the effect of GMFCS level on PODCI scores.

RESULTS: Global, Sports, Transfer, and Upper Extremity scores were different between all GMFCS levels ($p<0.05$) except for upper extremity scores which did not differ significantly between GMFCS levels II and III ($p=0.2$) (Table 1, Figures 1-4). Significant differences were not found for Happiness, Expectations, and Pain scores. Patients in GMFCS level IV had lower Global and physical function (Sports, Transfer, and Upper Extremity) scores compared with all other GMFCS levels. Scores for Happiness, Expectations, and Pain did not differ significantly between GMFCS IV and the other groups although the GMFCS IV group had the highest Expectations and Pain scores and the lowest Happiness scores.

Table 1: PODCI score distribution across GMFCS levels

	GMFCS I N=35	GMFCS II N=105	GMFCS III N=43	GMFCS IV N=31	P
Global	**76.4 (11.9)** ẋ‡Δ	**63.1 (14.7)** Ɨ‡Δ	**51.5 (15.5)** Ɨ ẋΔ	**38.2 (12.6)** Ɨ ẋ‡	**<0.0001**
Happiness	77.7 (21.4)	73.3 (22.4)	73.1 (28.3)	67.6 (32.2)	0.45
Expectations	83.3 (16.9)	82.0 (20.3)	84.8 (19.1)	87.4 (12.2)	0.55
Pain	71.4 (25.8)	70.5 (25.6)	72.7 (24.2)	76.6 (20.6)	0.67
Sports	**66.0 (14.8)** ẋ‡Δ	**46.3 (16.2)** Ɨ‡Δ	**30.1 (15.2)** Ɨ‡Δ	**19.3 (12.3)** Ɨ ẋ‡	**<0.0001**
Transfer	**90.0 (9.7)** ẋ‡Δ	**74.2 (16.3)** Ɨ‡Δ	**50.5 (19.0)** Ɨ ẋΔ	**23.9 (19.4)** Ɨ ẋ‡	**<0.0001**
Upper Extremity	**78.2 (14.4)** ẋ‡Δ	**61.3 (23.7)** ƗΔ	**52.6 (23.6)** ƗΔ	**33.0 (25.0)** Ɨ ẋ‡	**<0.0001**

Scores are reported as mean (SD). Symbols refer to differences between groups, scores in bold represent statistical significance ($p<0.05$) (Ɨ different than I, ẋ different than II, ‡ different than III, Δ different than IV).

DISCUSSION: The utilization of self/parent reported outcome measures allows for more patient/family centered assessments to ensure all of their concerns are being addressed. When these measures are examined alongside functional classification systems, they allow for a more complete clinical picture of the patient, improving treatment recommendations and subsequent outcomes. We found similar results to those in Barnes 2008 study [1], who found that PODCI global, transfers and sports scores were related to GMFCS levels for patients level I-III. However, we included patients GMFCS level IV in our analysis and found that PODCI upper extremity scores were significantly different between level IV and all other GMFCS levels although patients GMFCS levels II and III did not differ from each other in upper extremity scores. Interestingly, while patients in GMFCS level IV clearly had lower functional abilities than patients at lower GMFCS levels, there was much less difference in psychological outcomes like pain perception, happiness, and expectations. This highlights that functional outcomes and psychological outcomes can often be independent from one another. Examining the differences between GMFCS levels for PODCI individual scores allows for health care providers to intervene in areas that are of meaning to these patients and their families.

Figure 1

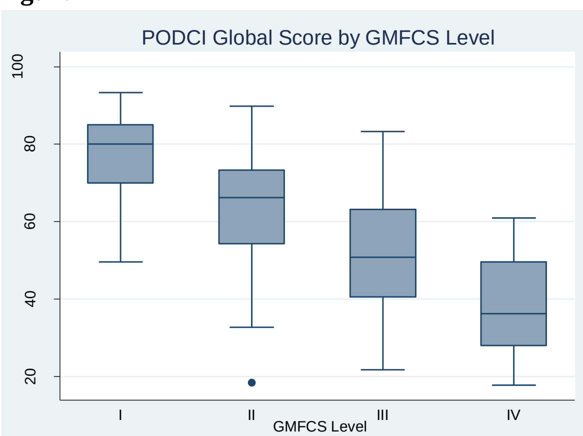


Figure 2

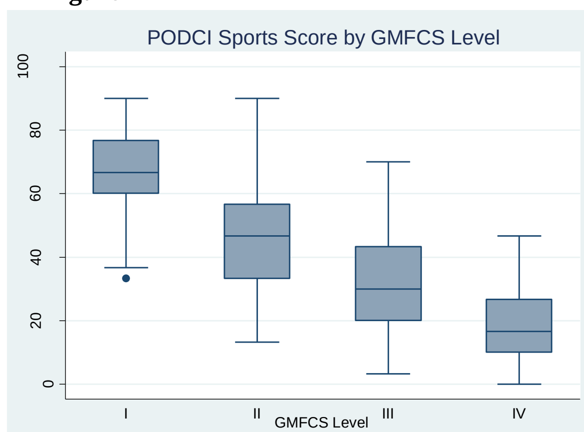


Figure 3

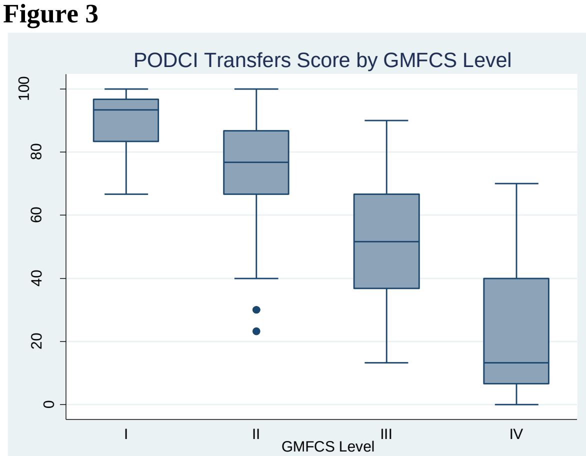
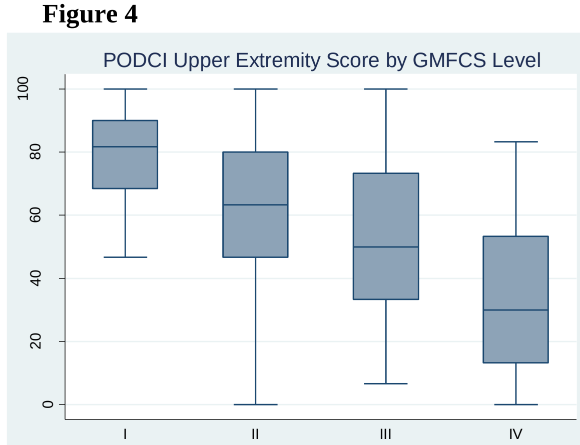


Figure 4



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UTILIZATION OF ACTIVITY MONITORING AND THE GOAL QUESTIONNAIRE TO ENHANCE TREATMENT RECOMMENDATIONS: A CASE STUDY.

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PATIENT HISTORY/INTRODUCTION
CM is a 13 year 5 month-old boy with a diagnosis of spastic diplegia cerebral palsy (CP), GMFCS II who was referred for 3-D gait analysis 7.5 months status-post bilateral talonavicular joint arthrodesis, and 2 years post previous SEMLS. Rotational alignment, crouch, and other soft tissue concerns were resolved. Primary family concerns include balance and stamina. He falls multiple times per day. Family priorities are focused on the Gait Function and Mobility domain of the Gait Outcomes Assessment List (GOAL) questionnaire after his recent surgery. Limited objective information regarding day-to-day activity was available to understand function and stamina concerns in the community. This focused case study highlights how the addition of activity monitoring data, assessment of functional decline and knowledge of family priorities can enhance and tailor treatment recommendations.

CLINICAL DATA:
Physical examination revealed generalized weakness, with less than antigravity strength of hip extensors, abductors, and foot/ankle musculature. Barefoot energy expenditure was 2.4x higher than normal with a decreased walking velocity. Total standardized GOAL scores decreased from the 82nd to the 46th percentile since his recent surgery. Reductions were noted in the Gait Function & Mobility, Pain Discomfort & Fatigue, and Physical Activity Sports & Recreation domains.

MOTION DATA:
Data were collected with a 12-camera motion capture system (120 Hz; Vicon, Oxford UK) with functional model calibration. Use of bilateral solid AFOs restored a heelstrike initial contact bilaterally, left swing phase knee flexion, and improved step length, stride length, velocity and stance time percentage bilaterally. (Figure 1). The Gait Deviation Index (GDI) improved from 69.1 to 76.6 (left) with minimal change on the right.

TREATMENT DECISIONS AND INDICATIONS:
Gait analysis data, energy expenditure, physical examination results and the GOAL questionnaire responses were consistent with recommendations for continued use of bilateral solid AFOS, core and foot/ankle strengthening, and promoting physical fitness activities. No further surgery was recommended. Activity monitoring was recommended to better understand home and community walking performance and activity due to concerns about decreased stamina before final recommendations.

OUTCOME: Eight consecutive days of at-home/community activity monitoring was collected (StepWatch 4). Capability of walking more steps than typical for children who function at GMFCS II was demonstrated; activity was less on weekends. The intensity of walking was categorized as 'low'(<15 steps/minute). (Figure 2).

Figure 1:

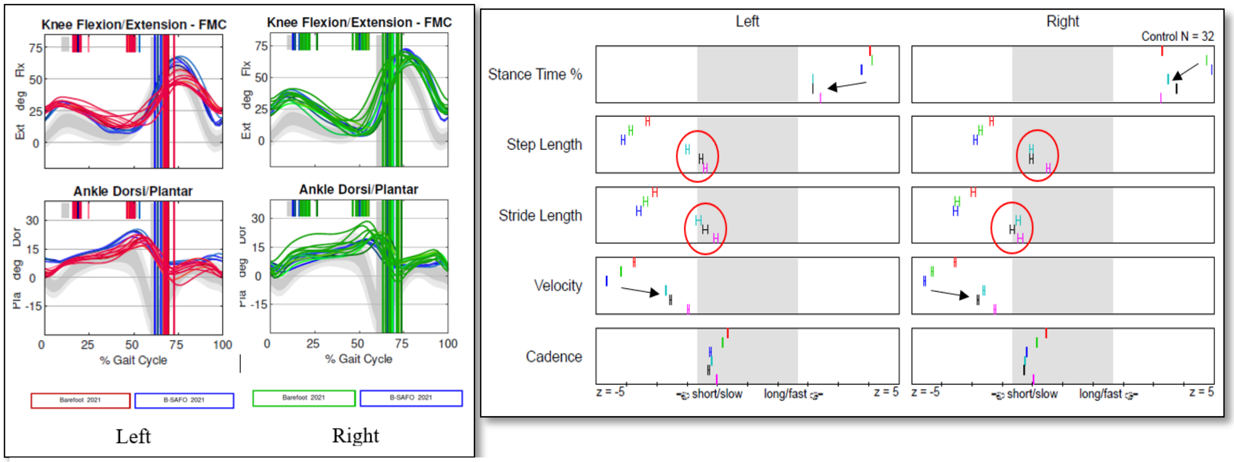

Figure 1: Left: Barefoot (red, green) and solid AFOs (blue) vs. control (gray) consistency data. 5 trials displayed. FMC=functional model calibration. Greater improvements are noted on left vs. right. **Right:** Linear parameter z-scores (individual trials in colors vs. control (gray 95% confidence interval)). Top 3 hash marks represent barefoot data; bottom 3 represent orthotic data. Cadence is unchanged, but step length, stride length, velocity and stance time percentage are all improved with orthotics (circled data and arrows highlight orthotic data).

Figure 2: Activity Monitoring Data

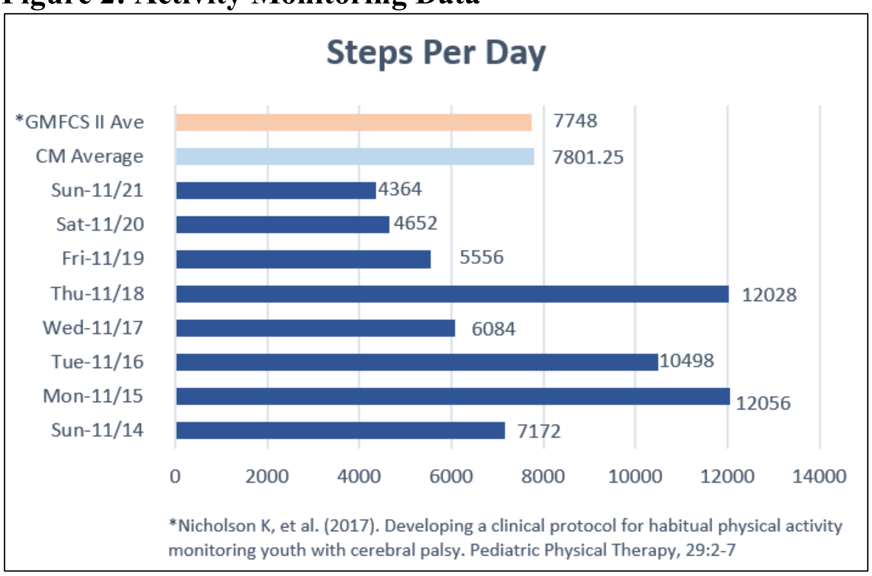

Figure 2: CM's average steps per day are consistent with GMFCS II. He is capable of walking greater than average steps per day for GMFCS II but typically walks less on the weekends.

SUMMARY: Activity monitoring data was consistent with specific item difficulties and priorities on the GOAL. Expanded treatment recommendations included focus on higher intensity activity, walking at faster speeds, and increasing weekend activity levels. Combining activity monitoring and the GOAL integrates objective community-based function data, family priorities, and walking capacity measures which enhances treatment recommendations made from clinical gait analysis data alone.

DISCLOSURE STATEMENT: The authors have nothing to disclose.

A multicenter study classifying transverse plane gait deviations of children with cerebral palsy into clinically meaningful biomechanical phenotypes using a novel algorithm

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INTRODUCTION

Identifying the gait pattern used by patients with cerebral palsy (CP) is a useful part of the treatment decision making process.[1-4] Clinical gait analysis provides quantitative information that can be used to quantitatively categorize gait patterns.[5] The objective of this work was to report on the prevalence of clinically meaningful transverse plane gait phenotypes in children with CP across four motion analysis centers in the United States using a novel algorithm.

CLINICAL SIGNIFICANCE

Deriving these clinically meaningful biomechanical phenotypes can provide objective classifications helpful in prescribing interventions.

METHODS

Data was used from 700 subjects that had completed instrumented gait analysis between January 1, 2019 and December 31, 2020. Discrete kinematic variables for the hip, knee, and ankle were extracted from a shared registry. Classification categories were developed based on definitions from literature and physician review (Figure 1). Classification categories were developed based on physician experience and review using a combination of the cutoff values in Table 1 and the algorithm displayed as a decision matrix flowchart in Figure 1. For transverse plane gait deviations, the classification begins based on foot progression angle (FPA). Transverse classifications included internal and external rotation levels 0, 1, 2, and 3 based on number of segments involved (0 means only the foot is internal/external in the transverse plane, but there is no involvement from the pelvis, hip or knee). Outlier data were removed based on averaged motion parameters collected from typically developing children at the 4 participating labs.

Table 1. Cut-off values for gait parameters

Gait Pattern	Internal	External
Mean Foot Progression Angle (Stance Phase)	> 6.3°	≤ -21.5°
Mean Knee Rotation (Stance Phase)	> 17.4°	≤ -15.8°
Mean Hip Rotation	> 19.0°	≤ -17.4°
Mean Pelvic Rotation	> 7.8°	≤ -7.3°

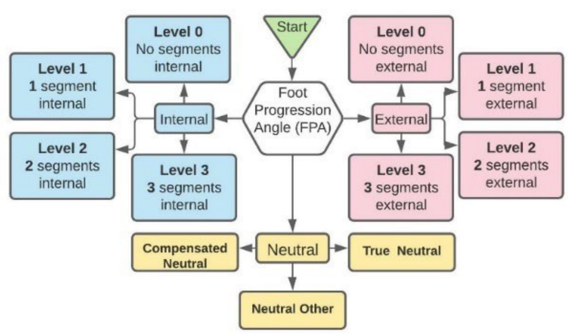

Figure 1. A novel algorithm for transverse plane gait classifications displayed as a flow chart. Note that for all levels of Internal and External, and for both Compensated Neutral and Other Neutral, the foot itself may have Adductus or Abductus. A one segment foot model cannot measure forefoot to hindfoot position.

RESULTS

Regardless of GMFCS level, the most common transverse plane phenotype was All Neutral (Table 2). All Neutral implies no transverse plane abnormalities in any lower body segment. In GMFCS I, Compensated Neutral (FPA is normal, but there is rotation in other segments of 10°

or less) was most common (16.1%). Internal 0 (only foot involvement, 18.5%) was most common in GMFCS II. GMFCS III showed an even split among External 0, Internal 0 and Internal 1 (14% each). GMFCS IV showed an even split among Internal I and Internal 0 (18% each). There was less internal FPA in GMFCS I than other GMFCS levels. There was more external FPA in GMFCS III and IV than I and II.

Table 2. Transverse plane classifications by GMFCS level

Transverse Plane Gait Deviation	GMFCS I n=1491	GMFCS II n=2057	GMFCS III n=892	GMFCS IV n=90	Total Population
Percent of Total Population	32.5%	44.8%	19.4%	2.0%	100.0%
All Neutral	46.3%	34.5%	23.2%	21.1%	35.7%
Compensated Neutral	16.1%	12.3%	10.9%	7.8%	13.1%
Neutral Other	12.1%	9.6%	11.9%	13.3%	11.1%
External 0	3.6%	6.5%	14.9%	14.4%	7.5%
External 1	4.5%	5.9%	7.6%	5.6%	5.8%
External 2	0.5%	0.9%	1.5%	0.0%	0.8%
External 3	0.0%	0.0%	0.0%	0.0%	0.0%
Internal 0	10.4%	18.5%	14.5%	17.8%	15.0%
Internal 1	6.1%	11.2%	14.1%	18.9%	10.3%
Internal 2	0.3%	0.6%	1.5%	1.1%	0.7%
Internal 3	0.1%	0.0%	0.0%	0.0%	0.0%

DISCUSSION

Deriving clinically meaningful biomechanical phenotypes can provide objective classifications to prescribe interventions. This novel algorithm could allow for automating the classification process into which would lead to improved efficiency and repeatability. This work demonstrates the ability of the algorithm to be applied across multiple centers. Future work aims to expand this algorithm to other joint measures, and to refine the algorithm beyond use of discrete data points.

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DISCLOSURE STATEMENT

Dr. Davids is a consultant to Orthopediatrics Inc. All other authors have nothing to disclose.

DOES STEP LENGTH INCREASE FOLLOWING SURGICAL CORRECTION FOR EQUINUS WITH A CONCOMITANT MEDIAL HAMSTRING LENGTHENING IN CHILDREN WITH HEMIPLEGIA?

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INTRODUCTION

Equinus contractures are the most common deformity in children with spastic cerebral palsy [1]. Hemiplegic patients often develop this equinus position on their involved side due to the imbalance of muscle tone and/or spasticity [2]. Plantarflexion contractures can lead to numerous variations in the subject's gait pattern. When initial contact is made with an equinus foot, full knee extension is often not achieved, and the resultant step length is shortened [3]. Surgical correction of an equinus contracture is most commonly performed via a tendo achilles lengthening (TAL) or by a gastrocnemius resection (GSR), depending upon the contributions to the equinus by the two plantarflexor muscles [4]. When the physical assessment reveals early limitations in knee extension or increased popliteal angles, a medial hamstring lengthening (MHL) may be concomitantly performed with the goals of improving knee extension in terminal swing and into initial contact. The purpose of this retrospective study is to determine if the addition of MHL to TAL/GSR for equinus contracture results in an increase in step length of the involved limb and improvements in temporal parameters of gait.

CLINICAL SIGNIFICANCE

In patients with hemiplegic cerebral palsy with an equinus contracture, surgical considerations for performing isolated TAL/GSR versus the addition of MHL should include which procedure may more positively impact the efficiency and velocity of gait. Objective data collected during 3D gait analysis provides measures to compare pre- and post- surgical information regarding efficacy of the procedure.

METHODS

Pre-operative and one-year post-operative 3D gait analysis data were retrospectively reviewed for a cohort of fifteen hemiplegic patients with identified equinus contractures on their involved side. Kinematic data were collected with a Vicon motion capture system (Vicon, CO) and five ATMI force plates (ATMI, NY) while the child walked at a self-selected speed. The plug-in-gait model was applied and data was processed using Vicon Nexus. A representative trial was selected and used for analysis of specific gait and cadence parameters. Statistical analysis of the change between pre- and post-operative visits for the TAL/GSR group and for the MHL group were conducted using paired t-tests ($p<0.05$).

RESULTS

The study cohort consisted of fifteen patients: twelve of which underwent isolated equinus correction, while three also underwent concomitant MHL. Of the twelve who underwent lengthening at the ankle only, ten had TAL and two had GSR; nine were GMFCS level I and three were GMFCS level II (pre-operatively). Of the three who underwent additional lengthening at the knee, two had TAL and one had GSR; two were GMFCS level I and one was

GMFCS level II (pre-operatively). The mean age for all participants pre-operatively was 8.2 years (range of 5.4-17.0 years). The post-operative visit occurred at an average of 1.5 years after the pre-operative visit (range of 1.2-2.0 years).

	TAL/GSR Change	w/ MHL Change	P value
Temporal Parameters:			
Cadence (% age matched TD)	-0.1 ± 10.1	8.7 ± 11.8	0.324
Walking Speed (% age matched TD)	6.9 ± 16.0	11.1 ± 13.6	0.667
Step Length (% age matched TD)	4.2 ± 10.6	1.3 ± 8.5	0.639
Knee:			
Minimum Flexion in SLS (degrees)	1.3 ± 11.9	-15.2 ± 7.6	0.033*
Minimum Flexion at IC (degrees)	-3.8 ± 6.4	-11.1 ± 5.6	0.130
Ankle:			
Maximum Dorsiflexion in SLS (degrees)	30.5 ± 20.1	39.7 ± 37.7	0.718
Dorsiflexion at IC (degrees)	14.4 ± 21.9	27.0 ± 25.2	0.723

Table 1. Comparison of change, pre-operative to post-operative, for each cohort *p<0.05
Abbreviations: SLS=Single limb support; IC=Initial contact; TD=typically developing

No significant differences were found in the temporal parameters between groups; this includes at the initial pre-operative examination, as well as after surgical correction. Kinematic data revealed improvement in the dynamic ankle and knee joint range of motion from pre- to post-, where the specific lengthenings were performed. This improved range of motion is reflected in the gait cycle during single limb stance and initial contact. Minimum knee flexion in the stance phase was the only variable to have statistical significance between the two cohorts. Limited sample size, especially for the MHL cohort, likely contributed to the lack of additional statistically significant findings.

DISCUSSION

This preliminary study was initiated by a physician inquiry and was started with a small sample size, which limits the data available for analysis and for interpretation of results. While results did not show a significant difference in any of the temporal parameters for either group, an increased number of patients should be gathered for the MHL group to verify the findings from the current n=3 population. The lengthening of the plantarflexors in all groups resulted in anticipated dorsiflexion during stance phase and initial contact. The added lengthening of the hamstrings resulted in improved knee extension in those phases as well. Improved functional range of motion of the ankle and knee, throughout the gait cycle, is the most significant benefit of either procedure.

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DISCLOSURE STATEMENT

The authors have no conflicts of interest to disclose.

ORTHOPAEDIC SURGICAL DECISION-MAKING AND EVALUATION: A CASE STUDY FOR A CHILD WITH CEREBRAL PALSY

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PATIENT HISTORY

Patient was a 6+11 year-old female, GMFCS II, diagnosed with acute atrophic thoracic spinal paralysis (T3-T7 atrophy) with associated spastic diplegia and bilateral hip dysplasia. . Developmental milestones were delayed with independent walking between 2.5-3 years. Previous surgery included bilateral proximal femoral varus derotational and pelvic osteotomies at 4 years (hardware removal 1 year post) Parent concerns included increased incidence of falling in all environments (home, school and community) multiple times a day and fatigue. The patient had a comprehensive 3D gait analysis at 5+1 and 6+11 years followed by additional orthopedic surgery and a post-surgery test at 8+4 years.

CLINICAL DATA

Clinical exam findings for pre post-surgery are summarized (Table 1). The selective Control Assessment of the Lower Extremity (SCALE) scores: were 2 right, 5 left pre-surgery.

	Pre-Surgery (6+11 yrs.)		**Post-Surgery (8+4 yrs.)**	
Height (cm)/Weight (kg)/BMI (kg/m^2)	117 / 21.0 / 15.3		121 / 25.3 / 17.3	
Walking Velocity (m/s)	0.90-1.02		1.11-1.19	
Side	Right	Left	Right	Left
Gait Deviation Index (GDI)	48-51	56-59	76-77	76-78
Knee extension strength	4	5	4	4
Ankle plantar flexion strength	U	2	1	2
Knee extension (deg)	5	5	5	10
Popliteal angle (deg)	-40	-45	-40	-40
Ankle dorsiflexion (knee 0 deg) (deg)	0	5	10	10
Ankle dorsiflexion (knee 90 deg) (deg)	5	10	15	20
Ely test (R/L)	1	1	2	2
Tibial torsion, foot thigh (deg)	5, -10	10, 5	15, 25	15, 20
Fem Ant, Int hip, Ext hip (deg)	30, 40, 65	35, 40, 65	10, 40, 60	15, 45, 60

Legend: Fem ant, femoral anteversion; Int hip, passive internal hip rotation; Ext hip, passive external hip rotation; yrs., years; U=unable.

MOTION DATA

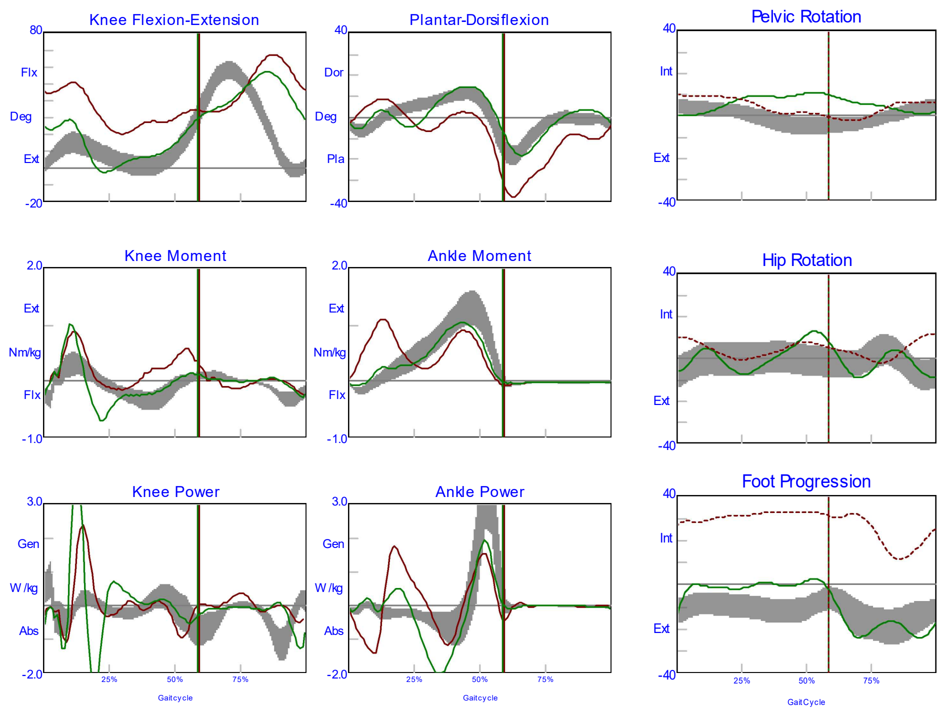
At pre-op, motion analysis data showed bilateral increased knee flexion at initial contact and in stance, increased equinus in mid-stance/mid-swing, and internal foot progression (Fig. 1). Kinematic and kinetic data were very consistent stride to stride (not shown).

TREATMENT DECISIONS AND INDICATIONS

1) Right tibial derotational osteotomy. Indications: right internal tibial torsion on clinical exam, internal foot progressing with neutral hip and pelvis rotation. Goal: improved prepositioning at initial contact, stance phase stability and swing phase clearance (catching right foot on left stance limb).
2) Bilateral gastrocnemius lengthening. Indications: plantar flexor spasticity and contracture, increased equinus in mid-swing and premature plantar flexion in mid stance.

Goal: improved stance phase stability and swing phase clearance.

3) Bilateral medial hamstring lengthenings. Indications: bilateral increased knee flexion at initial contact and in stance, bilateral hamstring spasticity on clinical exam. Goal: improved prepositioning at initial contact.
4) Gait decline between gait analysis tests completed at age 5 and 6 years, similar gait deviations including toe walking with AFO's (data not shown).
5) No left tibial derotation osteotomy: lesser deformity on clinical exam and gait.
6) Bilateral rectus femoris lengthening recommended/not performed. Indications: positive Ely test, rectus femoris activity in mid-swing, delayed peak knee flexion in swing, hamstring procedure, clearance problems in swing: Goal: improved clearance in swing.



a) b)

Figure 1: Pre (red/dashed) vs. post (green/solid) gait data for a) right side sagittal plane ankle and knee kinematics/kinetics and b) right side transverse plane kinematics.

TREATMENT OUTCOMES

Following surgery, the patient showed improved bilateral knee and ankle kinematics and kinetics (right side only shown) and right foot progression (Fig. 1). Pelvic tilt showed no change and hips showed increased extension (not shown) due to knee changes.

SUMMARY

The patient showed compromised prerequisites of typical gait including stance phase stability, swing phase clearance, prepositioning at initial contact that were improved post-surgery. The gait analysis clarified that the bilateral toe walking was a result of both knee and ankle kinematics. Evaluation of treatment outcomes with gait analysis data allowed a detailed understanding of which treatments impacted specific prerequisites of gait.

DISCLOSURE STATEMENT

The authors have no conflicts of interest to disclose.

QUANTIFYING TIBIOFEMORAL JOINT FORCES USING A SUBJECT-SPECIFIC VIRTUAL SIMULATOR REPRODUCING EXOSKELETAL-ASSISTED WALKING

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INTRODUCTION

Robotic exoskeletal devices have vast, but largely untapped, potential to restore upright mobility to individuals with neurological disorders, including those of spinal cord injury (SCI), stroke, traumatic brain injury, and multiple sclerosis. The growing demand for these devices necessitates the development of technology to systematically characterize human and robot interactions during exoskeletal-assisted walking (EAW). However, such parametric studies are cost-prohibitive and not feasible using experimental methods. For instance, tibiofemoral (TF) joint forces provide an accurate measure of bone loading during EAW, but experimental methods to quantify TF joint forces are invasive and currently not feasible. Hence, computational simulation is the only viable alternative. Accordingly, the goal of this study was to develop a computational framework to quantify TF joint forces using a subject-specific virtual simulator that reproduces EAW.

CLINICAL SIGNIFICANCE

Persons with chronic SCI are at increased risk of fractures during EAW due to severe sublesional osteoporosis. Quantification of TF forces during EAW will provide much-needed understanding of the loading forces applied to the long bones of the legs, which will assist clinicians to minimize fractures during EAW.

METHODS

An able-bodied man (41 years, 178 cm, 86.2 kg) was recruited for this study. Prior to participation, the subject provided written informed consent. The subject was trained in a ReWalk P6.0, and his 3-D motion was analyzed during unassisted and exoskeletal-assisted walking maneuvers. All 3-D motion data were collected from a single session and included simultaneous measurements of marker trajectories, ground reaction forces, electromyography (EMG), and exoskeleton motor angle and torque data.

We adapted a full-body OpenSim model [1] to simulate unassisted and exoskeletal-assisted walking maneuvers. A full-scale geometry of the ReWalk P6.0 was integrated with the musculoskeletal model (Figure 1A). The human-robot model had 16 degrees of freedom to represent the exoskeleton (two at the hips, two at the knees, six at each foot plate to interact with the human foot, and a kinematic constraint to anchor the two sides at pelvic band) in addition to the 37 degrees of freedom to represent the human. The human-robot model represented interaction using OpenSim's bushing forces (three linear spring-damper systems, two for lower legs, two for upper legs, and one at the torso) [2]. We scaled the generic musculoskeletal model to match the anthropometry of the subject. Inverse kinematics was performed to obtain joint angles, and inverse dynamics to obtain joint moments. Using OpenSim's force reporter [2], interaction forces were quantified.

EMG-tracked muscle-driven simulations of EAW were performed using OpenSim Moco [3] to compute TF joint forces. The DeGrooteFregly2016 [4] muscle model was used to reproduce motion at the lower extremity, and ideal torque actuators were employed to displace the torso and the upper body. In our simulations of EAW, our estimated exoskeleton interaction forces (Bushing Forces) were included and applied to a model without exoskeleton geometry. In addition to this approach, EAW was simulated using two other approaches, one with exoskeleton motor torques prescribed directly at each joint (Prescribed Torque) and the other excluding exoskeletal interaction forces or motor torques (w/o Interaction).

RESULTS

Our virtual simulator reproduced unassisted and exoskeletal-assisted walking maneuvers within acceptable ranges (average RMSE 1.2 cm for unassisted and 1.3 cm for EAW). Compressive TF forces obtained from unassisted walking and EAW were determined (Figures 1B and 1C). Peak compressive TF forces during unassisted walking were 2.75 BW and 2.80 BW for the left and right sides, respectively. Peak compressive TF forces during EAW were 2.76 BW and 3.54 BW (Bushing Forces) in comparison to 3.06 BW and 3.32 BW (w/o Interaction) and 5.02 BW and 5.55 BW (Prescribed Torque) for the left and right sides, respectively. In addition to compressive TF forces, TF shear and anterior-posterior forces were estimated (not shown).

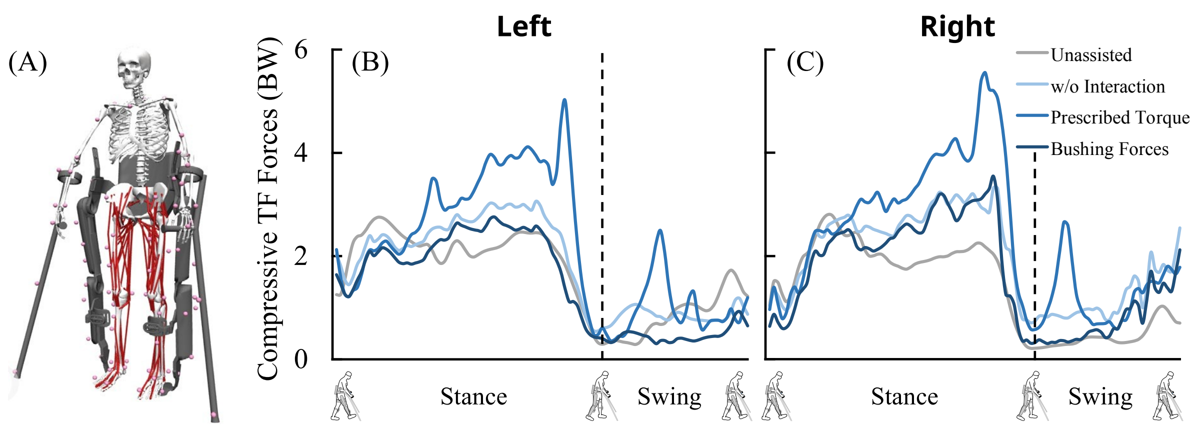

Figure 1: A subject-specific virtual simulator of EAW (A). Compressive TF forces from unassisted and EAW maneuvers for left and right sides (B, C).

DISCUSSION

Our framework provides a novel approach to quantify TF joint forces during EAW. The described virtual simulator is a low-risk and cost-effective alternative to invasive approaches to characterize human-robot interaction during EAW. Our noninvasive approach provides the groundwork for future parametric studies that address human-robot interactions.

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DISCLOSURE STATEMENT No conflicts of interest to disclose.

Estimating Ground Reaction Forces (GRF) and Lower Extremity Joint Moment in Multiple Walking Environment Using Deep Learning

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INTRODUCTION

Gait analyses provide useful information for assessing the performance of dynamic activities. Specifically, Ground Reaction Forces (GRFs) and lower extremity joint moments can provide valuable information to clinicians to assess the patients' walking performance associated with their treatments. Traditional techniques for measuring GRF and joint moment require strict lab setup (motion capture, force plate) and expert post-processing, making it difficult to assess these biomechanics parameters outside the lab. To address these issues, Inertial Measurement Unit (IMU) sensors with machine learning algorithms are suggested to estimate GRFs and joint moments [1]. However, these studies were confined to a highly repetitive motion (treadmill and/or level-ground). Also, we are unclear of their algorithm's capability for reliable assessments of GRF and joint moment in various walking conditions, including stairs and slope walking. Thus, we propose a novel deep learning algorithm that enables an accurate real-time estimate of GRF and joint moment in various walking conditions using IMU sensors.

CLINICAL SIGNIFICANCE

Estimating GRF and joint moments in various walking conditions outside the lab is essential to assist clinicians to assess and treat patients' pathologic walking functions.

METHODS

We leveraged a publicly available dataset to validate our algorithm [2]. The dataset contains multiple walking scenarios (i.e., treadmill, level-ground, slope, and stair walking) of 20 participants. We applied segmentation (gait cycle extraction) on the dataset to get GRFs and corresponding joint moments. As IMU sensors were only available for the right side of the leg, we estimated GRFs and joint moments for the right leg only. We used three IMU sensors at the thigh, foot, and shank as inputs to predict 3D GRF and joint moment of the hip, knee, and ankle. We used kinetics-Net (FM) for the deep learning model as demonstrated in Figure 1. In the Kinetics-Net (FM), we utilized different layers, i.e., GRU, Conv1D, Conv2D, Fully Connected (FC), to construct three subnetworks (GRU-Net, GRU-Conv1D-Net, GRU-Conv2D-Net). The reason for building subnetworks is to introduce diversity in the prediction and improve the performance using the ensemble method. These subnetworks are ensembled end-to-end to achieve better performance than each subnetwork separately. We also designed a Fusion Module (FM) to combine output from three subnetworks. The purpose of FM is to put proper weight to the subnetworks to optimize the performance of the Kinetics-Net, which in return perform better than a simple model averaging. We compared the results of Kinetics-Net (FM) with a simple model averaging approach (Kinetics-Net (A)) in the demonstration section. Figure 1 shows our proposed deep learning model Kinetics-Net (FM) with primary building blocks.

DEMONSTRATION

We used Normalized Root Mean Square Error (NRMSE) and Pearson Correlation Coefficient (PCC) as evaluation metrics for the proposed method. In Table 1, the mean and standard deviation of NRMSE and PCC from Kinetics-Net (FM), Kinetics-Net (A) are shown for all the participants in all walking conditions. Kinetics-Net (FM) outperforms Kinetics-Net (A) for all joint moment and GRF, which demonstrates the effectiveness of the fusion module over a simple average approach for the ensemble of the models.

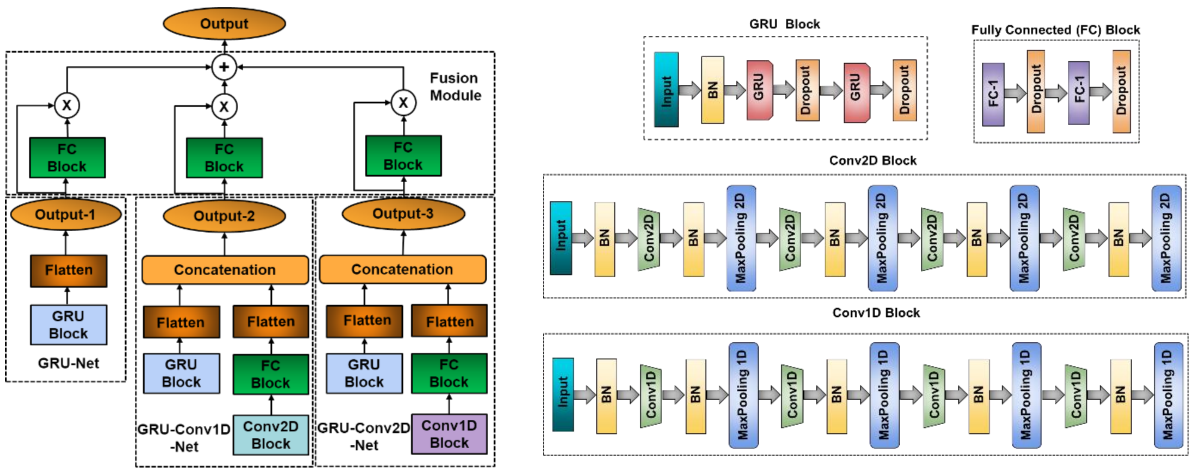

Figure 1: Kinetics-Net (FM) (left), the primary building blocks of Kinetics-Net (FM) (right)

Table 1: Mean± SD of NRMSE and PCC for Kinetics-Net (A) and Kinetics-Net (FM)

Metric	Model	Hip moment	Knee moment	Ankle moment	Mediola-teral GRF	Vertical GRF	Anterior-posterior GRF	Mean
NRMSE	Kinetics-Net (A)	4.91±1.19	4.65±1.52	4.49±2.64	4.69±1.56	5.00±2.31	3.38±0.65	4.52±0.92
	Kinetics-Net (FM)	**4.74±1.18**	**4.31±1.30**	**4.20±2.26**	**4.56±1.50**	**4.97±2.38**	**3.34±0.70**	**4.35±0.81**
PCC	Kinetics-Net (A)	0.895±0.037	0.879±0.055	0.938±0.035	0.861±0.054	0.975±0.029	0.951±0.026	0.916±0.033
	Kinetics-Net (FM)	**0.901±0.030**	**0.890±0.043**	**0.941±0.038**	**0.867±0.046**	**0.975±0.029**	**0.952±0.028**	**0.921±0.030**

SUMMARY

Our end-to-end trained ensemble model with a fusion model can reliably predict lower extremity joint moments and GRFs using three IMU sensors in different walking conditions. Our estimated moments and GRFs can be helpful to monitor patients without any constrained lab environment and expertise in data processing.

ACKNOWLEDGMENTS

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DISCLOSURE STATEMENT

The authors have no conflicts of interest to disclose.

ESTIMATION of MUSCLTENDON PARAMETERS USING OPTIMIZATION ALGORITHM

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INTRODUCTION

Short and tight muscle characteristics are common for people with movement disabilities such as stroke survivors and children with cerebral palsy. Passive musculotendon parameters (PMPs) (i.e., slack length and stiffness) play an important role in walking functions for people with movement disabilities. Current musculoskeletal modeling studies use a generic musculoskeletal model made with unimpaired individuals' PMPs. These parameters are not subject-specific and may result in discrepancies between experimental results and computational analyses.

Although shear wave ultrasound can be used to measure the PMPs, it can only quantify stiffness indirectly through shear wave intensity. The method only measures selected regions of, not whole musculotendon units [1] limited to apply these outcomes to computational musculoskeletal modeling analyses. Previous attempts to estimate the PMPs using optimization algorithms have been inaccurate because of their prioritization toward active parameters [2].

CLINICAL SIGNIFICANCE

We present a new method that enables comprehensive and accurate musculotendon slack length (MSL) and muscle stiffness (MS) estimations using an optimal control optimization.

METHODS

We used the PMPs of a generic musculoskeletal model developed by Rajagopal et al. [3] as a ground truth to evaluate the accuracy of our methods. First, we positioned the musculoskeletal model in a sitting position for a low-speed isokinetic knee flexion (5 °/sec) from 0° to 110° with the hip and ankle angles locked at 90° and 0° respectively. To consider the knee joint torques derived only from passive musculotendon forces and gravity, we deactivate the muscles. We applied an external actuator onto the knee joint to achieve the low-speed isokinetic motion. To control the external actuator, we used MOCO [4] to minimize the difference between the desired and simulated knee kinematics and summation of actuator's input.

To calculate the MS and MSL, we used the Millard et al. [5] musculotendon passive parameter equation $f_0^M\left(f^{PE}(\tilde{l}^M)\cos\alpha - f^T(\tilde{l}^T)\right) = 0$. Where $f_0^M$is the maximum muscle force at optimal fiber length and $\tilde{l}^M$ is normalized fiber length, and $\tilde{l}^T$ is the tendon length normalized by the MSL. The calculated passive musculotendon forces (f^{PE}, f^T) will be combined with given muscle moment arms, body segment geometries, and optimal fiber lengths to calculate knee joint torque. We arbitrarily changed the MSs and MSLs in the model and then used OpenSim MOCO to find the optimal MSLs and MSs that minimize error between the reference and derived knee motion. To validate the outcome of the optimization algorithm, we used $\frac{P_r - P_e}{P_r} * 100$ to calculate error. Where, P_r is the ground truth PMPs and P_e the estimated PMPs from the optimization algorithm.

DEMONSTRATION

The joint angle error between the reference knee motion data and the motion derived with the model using the estimated PMPs is less than 0.5 °. This suggests that the entire estimated PMPs were accurately operates knee motion (Figure 1). The error in MSL estimation was less than 3% while the estimation error for MS was less than 12% (Table 1). Bicep femoris short head had the largest error for both MSL and MS. This might be due to the smallest maximum isometric force compared to the other muscles, leading a small contribution to the net knee torque. On the other hand, the semimembranosus and semitendinosus had the most accurate estimations. Because of the initial hip position (90°) with full knee extension, these muscles dominantly produced passive forces form a large amount of stretching. As a result, the algorithm is more sensitive to tune PMPs of these muscles.

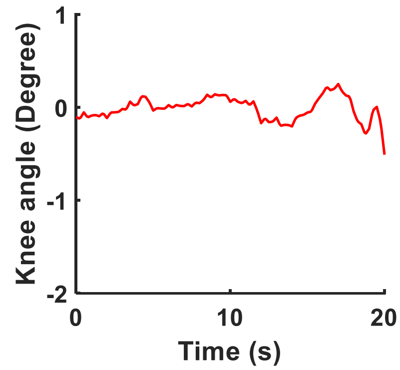

Figure 1. Error between the reference slow isokinetic knee flexion and the calculated knee angle using the derived PMPs.

Table 1: Error of estimated muscle slack length and stiffness of biceps femoris long head (bflh), biceps femoris short head (bfsh), gastrocnemius lateralis (glat), gastrocnemius medialis (gmed), rectus femoris(rfem), semimembranosus (smem), semitendinosus (sten), vastus intermedius (vasint), vastus lateralis (vaslat), and vastus medialis (vmed). Negative error and positive error represent underestimate and overestimate respectively.

Error (%)	**bflh**	**bfsh**	**glat**	**gmed**	**rfem**	**smem**	**sten**	**vasint**	**vaslat**	**vmed**
MSL	0.20	-2.5	-1.1	-1.5	1.6	0.2	0.4	-0.9	-1.1	-0.8
MS	-1.07	11.4	10.5	5.8	9.6	-1.4	-1.0	5.7	5.5	3.0

SUMMARY

In this paper, we suggested a new method that estimates PMPs with optimization algorithm during passive quasi-isokinetic condition. By arbitrarily altering the generic model PMPs and implementing our method to estimate their original value we verified our optimization algorithm. This validation of optimization algorithm and testing method will enable to accurately estimate PMP not only unimpaired individuals but also individuals with pathologic muscle characteristics.

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DISCLOSURE STATEMENT

Authors have no conflicts of interest to disclose.

Deriving Gastrocnemius Moment Arm Using Motion Capture, Ultrasound, and Musculoskeletal Model

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INTRODUCTION

A muscle's moment arm is a geometric characteristic that determines the transformation from a linear muscle force to a joint moment, it plays a significant role in assessing joint performance for athletes, elders, and people with movement disabilities. Magnetic resonance imaging (MRI) has previously been used to measure a muscle's moment arm. However, it is less accessible due to prohibitive cost, measurement volume constraints due to the narrow tunnel size, and requirements of trained experts [1]. This 3d ultrasound method (3DUS) could be an alternative that overcomes these challenges. Previous studies have been limited to calculating moment arm for a joint with a single degree of freedom and a simple line of musculotendon action [2]. This was due to limited image capture area and the complexity of calculating the moment arm with a multi-degree of freedom joint movement (i.e., knee joint), and a nonlinear muscle pathway. This is the first study to measure the moment arm of highly non-linear shape of the gastrocnemius muscle with a two degree of freedom knee joint using an ultrasound, a motion capture system, and musculoskeletal modeling.

CLINICAL SIGNIFICANCE

The proposed methodology accurately measures the gastrocnemius's moment arm about the knee, improving the accuracy and accessibility of musculoskeletal modeling. These models are useful in assessing pathologic muscle behavior and in the development of assistive devices.

METHODS

Our ultrasound method was applied to two young healthy males, and 57 kg 93 kg, respectively. Five reflective markers were placed onto the ultrasound probe to establish the ultrasound's coordinates and nine reflective markers were attached, following a modified Helen Hayes [3] marker set on the participant's leg to identify the joint angle and to reconstruct the shank and thigh segments with a motion capture system throughout the experiment. During the experiments, the participants laid on their side with their leg positioned at either 0°, 60°, 90°, or 110° of knee flexion. At each of these knee angles, their foot was positioned at 10°of dorsiflexion, 20° plantarflexion, and 0° of flexion. The ultrasound was then drawn distally from the knee to the ankle over the left lateral gastrocnemius. The image sequence produced by the ultrasound was processed in ImageJ to find the location of the muscle's centroid. The motion capture data arm.calculation was then interpolated to find the ultrasound's position when each ultrasound image was captured.

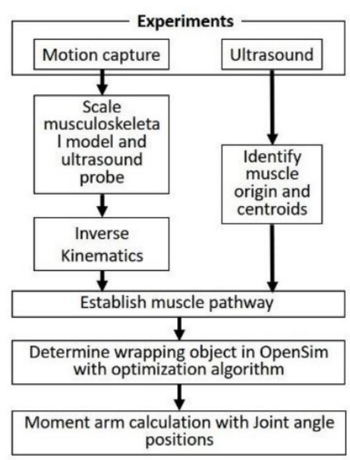

Figure 1: An overview of data processing from experimental collection to moment

Using a scaled Thelen model in OpenSim [4], the muscle centroid data was transformed from the 2D ultrasound image coordinates into the 3D coordinates of the motion capture environment, representing the position of the centerline of the muscle in reference to the origin of the tibia. To generate the muscle pathway in OpenSim, we created a wrapping object that minimized error between the measured pathway and the pathway in OpenSim using an optimization algorithm. Translation and rotation movement of the knee joint was also reported by the scaled musculoskeletal model. Using the established muscle pathway and joint movement, we calculated the muscle moment arm with OpenSim. The experimental moment arms were then compared with two scaled musculoskeletal models developed by Thelen et al. [4] and Rajagopal et al. [5]

DEMONSTRATION

Our moment arm results closely match with Rajagopal model when the knee is fully extended. However, as knee flexion angle increases, the Thelen model better matches our experimental moment arm. Changes in ankle angle do not affect knee moment arm of lateral gastrocnemius suggesting knee angle dominantly impact changes in muscle moment arm (Figure 2). As a result, the currently available generic musculoskeletal models do not accurately represent the moment arm of the lateral gastrocnemius of each participant.

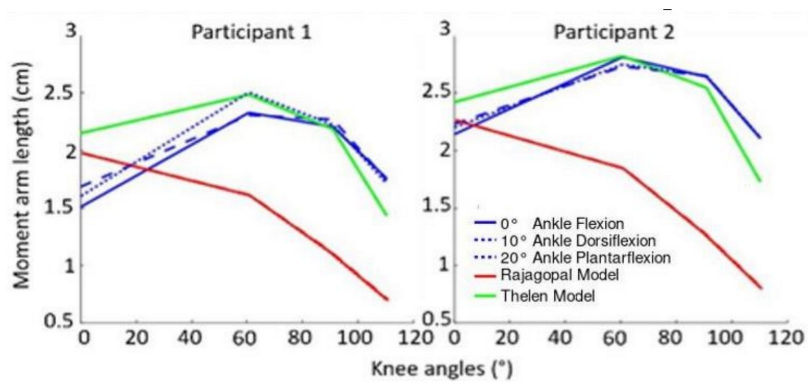

Figure 2: Comparisons of lateral gastrocnemius moment arms between experimental pathways and the moment arms of two scaled generic models

SUMMARY

Our new method enables for accurate moment arm measurement for muscles and joints that have complex geometry and movement function. This highlights the limitation of scaled generic models and the need for an updated muscle moment arm to match each experimental subject. To further develop this study subjects should be postured as to release or apply tension to the gastrocnemius, a larger sample size needs to be obtained, and more muscles need to be measured.

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DISCLOSURE STATEMENT

The authors have no conflicts of interest to disclose.

Electroymyography: What's What and The Interpretation of Data

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Purpose: This 1-hour session will allow participants to participate in case-based discussions to improve their skills in understanding kinesiologic EMG. This intermediate course is intended for individuals who have some experience with EMGs but would like to improve their understanding and skills in interpreting clinical EMG data. We will discuss EMG data quality issues that impact clinical interpretation. This course consists predominantly of mini-topics coupled with use-case-based discussions of EMGs in adult and pediatric clinical cases.

Intended audience: PTs, engineers, and kinesiologists. This is an intermediate-level course.

At the completion of this course, participants should be able to:

Discuss EMG data and identify physiologic and hardware related signal and noise

Develop clinical reasoning guided by EMG data in case-based scenarios

Interpret EMG within the context of complex clinical applications and cases

Prerequisite knowledge:

Basic knowledge of the underlying electrophysiology of muscle activity

Familiarity with surface EMG techniques and reporting

Basic familiarity with kinematics and kinetics of walking

Outline of course content:

5 minutes	Introduction and Session Format	
5-20 minutes	EMG Signals	Data quality issues. EMG Raw vs Linear Envelopes of EMG vs Frequency analysis Case examples & discussion
20-35	EMG Data forms	Tone, spasticity, amplitude/timing Case examples & discussion
35-50	Interpretation of Multi-channels in multi-segment movement	"Synergy" analysis Case examples & discussion
50-60	Discussion, Q & A	

KINEMATICS AND STRENGTH OUTCOMES FOLLOWING EARLY COMBINED NERVE AND TENDON TRANSFER VERSUS TENDON TRASFER ALONE FOR FOOT DROP AFTER MULTI-LIGAMENTOUS KNEE INJURY

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INTRODUCTION

Peroneal nerve palsy (also known as drop foot) can be a consequence of multi-ligamentous knee injuries (MLKI). Current treatment options include expectant management, nerve grafting, or tendon transfer. However, results are still mixed with less than half of patients experiencing meaningful functional recovery [1,2,3]. The early combination of partial tibial nerve transfer and posterior tibial tendon transfer may be a novel surgical strategy for management of this challenging clinical condition. The purpose of the case series was to compare the outcomes of early combined partial tibial nerve transfer and posterior tibial tendon and (NTT) to posterior tibial tendon transfer alone (TT) for the treatment of traumatic peroneal nerve palsy after unilateral MLKI at a minimum of two years postoperatively.

CLINICAL SIGNIFICANCE

Despite the small sample size, long term results of early combined nerve and tendon transfer were better than tendon transfer alone. Following NTT, a select group of individuals had improved ankle function and fewer concurrent compensatory motions during swing. This procedure may offer promise to these individuals following traumatic MLKI.

METHODS

Subjects: Following MLKI with unilateral drop foot, two males (NTT1, NTT2) underwent the combined procedure, one male (TT1) underwent tendon transfer alone. Subjects were tested at 5, 5, and 6 years post-operatively, respectively. The two NTT had surgery 6 months following trauma, while the traditional TT surgery was performed 14 years post injury. All subjects signed informed consent in this IRB approved study.

3D kinematics were collected (100fps) during level walking at self-selected speed for at least 5 cycles. Isokinetic dorsiflexor strength was measured for 5 trials at 60 deg/sec. Endurance was measured for 15 trials at 120 deg/sec (Biodex). All measures were compared between the affected and the unaffected side. Due to the limited sample size, no additional formal statistical analysis was performed across the groups.

RESULTS

During level walking, dorsiflexion was present on the affected side throughout swing and at initial contact only in the NTT individuals (Figure 1). For subject TT1, the affected ankle remained plantarflexed throughout swing phase. First rocker, eccentric ankle plantarflexion following initial contact, on the affected side was present only for the NTT subjects. Compensatory hip and knee flexion during swing on the affected side were present only in the TT subject. Hip and knee flexion were symmetrical for the two NTT individuals.

Ankle dorsiflexion strength and endurance were decreased compared to unaffected side in all subjects (Table 1). Only the TT subject was unable to generate torque on the affected side.

Table 1. Ankle dorsiflexion strength and endurance measures with Biodex machine

	NTT1	NTT2	TT1
Peak Torque (Nm)			
Affected side	17.5	11.7	0
Unaffected side	25.2	25.2	7.1
% Difference	**-30.7**	**-53.4**	**-100**

	NTT1	NTT2	TT1
Total work (J)			
Affected side	51.5	33.2	0
Unaffected side	108.1	111.4	19.2
% Difference	**-52.3**	**-70.2**	**-100**

DISCUSSION

In this small case series, following a novel combination of nerve and tendon transfer, long term measures of kinematics were more symmetrical than in tendon transfer alone. These findings are the first step in evaluating the effectively of the early combined nerve and tendon transfer. The nerve transfer permits re-innervation of the anterior tibialis and more effective active dorsiflexion. The tendon transfer alone may provide only tenodesis rather than active contraction. Although dorsiflexion is seen kinematically with NTT, weakness persists on strength testing. These results are limited due to the small sample size. Sample size was limited due to the number of individuals who were available at this time. As the popularity of this procedure increases, additional subjects will be available for long term testing. The combination of nerve and tendon transfer may offer promise to these individuals following traumatic MLKI.

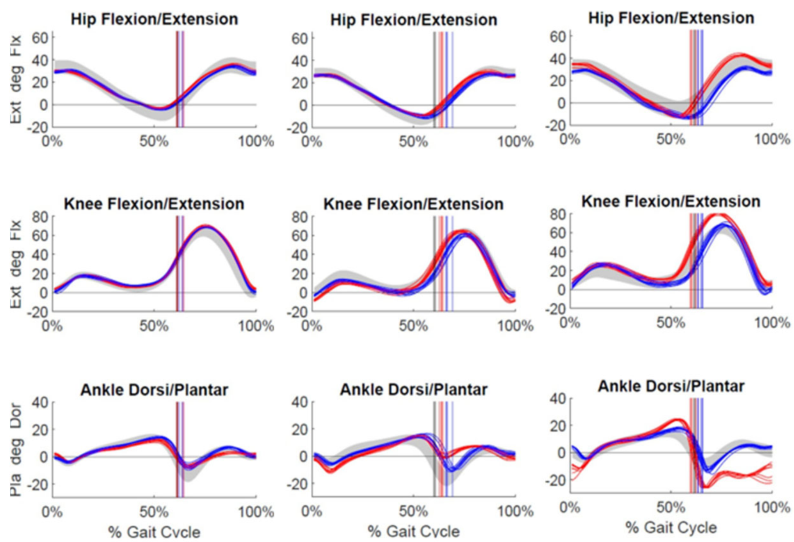

Figure 1. Kinematics. Affected (red) vs. Unaffected (blue)

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ACKNOWLEDGMENTS

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DISCLOSURE STATEMENT

The authors report no financial, consultative, institutional, and other relationships that might lead to a conflict of interest.

RELATIONSHIP BETWEEN STEREOGNOSIS AND ACTIVE RANGE OF MOTION FOR CHILDREN WITH HEMIPLEGIA CEREBRAL PALSY

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INTRODUCTION

Children with cerebral palsy (CP) experience a range of impairments including muscle weakness, sensory deficits, increased muscle spasticity, and upper limb dysfunction [1]. These impairments interfere with a child's ability to interact effectively in their environment and also decrease the child's daily self-care activities [2]. In the past, treatment for unilateral CP has had a strong focus on correcting motor impairments such as muscle weakness and/or spasticity [3]. Recent findings reported that the prevalence of sensory processing impairments in children with hemiplegia are up to 75% or more [4]. To date, there is minimal research assessing the relationship between active range of motion (AROM) and sensory processing (tactile discrimination) deficits in children with CP.

CLINICAL SIGNIFICANCE

The purpose of this study is to determine if there is a relationship between active AROM and tactile discrimination for children previously diagnosed with unilateral CP.

METHODS

This is an IRB approved retrospective study. Potential participants were previously evaluated for a clinical assessment of upper extremity functions in the Motion Analysis Center between August 2014 and July 2021. Inclusion criteria were: patients with the primary diagnosis of cerebral palsy, spastic hemiplegia, ages 7-21 years old at the time of the evaluation. Patients performed clinical assessment of active range of motion and stereognosis assessments during the visit. Exclusion criteria were patients who had undergone botulinum toxin injections in the past 6 months or surgical interventions in the past 12 months.

During the clinical assessment, each patient underwent a history and evaluation by a physical therapist and then performed active range of motion assessment with a three-dimensional motion analysis system. Reflective markers were placed on: left scapula (offset), bilateral lateral humeral heads, bilateral lateral mid-humerus, bilateral lateral humeral epicondyles, bilateral mid-forearms in line with thumb, bilateral radial styloid process, bilateral ulnar styloid process, and bilateral middle fingers proximal phalange bases and heads.

To assess active range of motion (AROM), the patients were seated on a height adjustable bench so both feet could be flat on the floor. Twelve 3-D motion capture cameras were used (Vicon Vantage Cameras, Nexus 2.12.1 software, Culver City, CA) for data collection and post processing. Three trials of active range of motion first with their uninvolved and then their involved arm were averaged. The motions performed were maximum wrist extension and flexion, and active forearm supination and pronation. After removal of reflective markers, regardless of which arm is more involved, patients performed stereognosis assessment first with their right upper extremity and then with their left. Stereognosis assessment consisted of patient sitting in a chair and then placing their hand on a bedside table with a small curtain to prevent the patient from seeing the objects and shapes. Five different objects and four different shapes were placed under the patient's hand by the therapist. Patients were provided a diagram of objects and shapes and were asked to identify the object under their hand. Patient then received a score of 0 (incorrect) or 1 (correct identification) of each object and shape. Patients were not allowed to see the object or shapes at any time. The therapist did not inform patient if their answers were correct or incorrect.

Statistical analysis were performed with SPSS v28 (IBM, Armonk, New York)**.** Pearson's Correlation coefficients between AROM and object scores and shape scores.

RESULTS

Sixty-two patients (26 females) met inclusion criteria. Majority (82%) of participants had Manual Ability Classification System (MACS) score of II, with 14 % score of I and 3 % score of III. Per family report, 50 of the patients performed activities of daily living with one hand.

On average, the uninvolved arms demonstrated more wrist and finger extension ($p<0.001$), and total excursion ($p<0.001$) compared to involved arms. Uninvolved arms demonstrated more pronation ($p=0.004$) and supination ($p<0.001$). Uninvolved arms demonstrated higher rates of correct identification of shapes and objects ($p<0.001$).

For the involved arms, object identification was moderately correlated with active wrist extension/flexion excursion ($r = .37$), and forearm pronation/supination excursion ($r = .32$). No relationship between object identification scores and finger flexion/extension excursion ($r = 0.19$). For the involved arms shape identification demonstrated strong correlation with active wrist extension/flexion excursion ($r = .48$). For the involved arms shape identification demonstrated weak correlation with active finger total excursion ($r = .28$) and forearm pronation/supination excursion ($r = .27$). No relationship between stereognosis scores and active range of motion was observed for the uninvolved arms.

DISCUSSION

Positive relationship was observed between stereognosis score and active wrist and finger excursion for the involved arms of children diagnosed with cerebral palsy, hemiplegia. Children with less AROM demonstrated a lower ability to identify objects and shapes correctly. This data implies that improving ROM of an involved upper extremity through treatment interventions could improve stereognosis ability for a patient with CP. Further clinical studies are need to confirm this theory.

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

DEVELOPMENT OF A FUNCTIONAL PERSONAL WORKSPACE IN CEREBRAL PALSY CHILDREN

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INTRODUCTION

Children with cerebral palsy (CP) demonstrate impairments that affect reaching and grasping tasks. These tasks are imperative for completing activities of daily living (ADL) such as personal hygiene, self-feeding, and dressing. A reachable workspace was developed for use in neuromuscular disorders (i.e. muscular dystrophy) that describes the maximum reachable area from the body [1]. However, this reachable workspace may not provide information related to ADLs related to self-care. Therefore, a functional personal workspace, focused on reachable space near the body, may be more pertinent to self-care ADLs.

CLINICAL SIGNIFICANCE

To our knowledge, there have been no personal workspaces developed using functional upper extremity motions for typically developing (TD) children and on children with hemiplegic cerebral palsy (CP).

METHODS

This is an IRB approved retrospective study. Fifty-three (24 TD, 29 CP) subjects had seven retroreflective markers placed on their upper body (bilateral ulnar styloid, bilateral humeral lateral epicondyle, bilateral lateral shoulder, posterior lateral offset).

Personal workspace trials started and ended with arms by participant's side. A single trial involved moving hands from the initial position to the top of the head, to the subject's mouth, to the subject's umbilicus, to the subject's back pocket, and back to initial position. Each personal workspace trial was separated into individual motions for post processing (hand to head [HtH], hand to mouth [HtM], hand to belly [HtB], and hand to pocket [HtP]) that were also analyzed separately. Subject's arms were analyzed individually. In TD children, dominant and non-dominant arms were identified. In CP subjects, the involved and uninvolved limbs were seperated.

Kinematic data were collected in Vicon Nexus v. 2.9.3 using 12 Vicon Vantage V16 cameras (Vicon Motion Systems, Oxford, UK). Data were processed in Matlab (Mathworks, Natick, MA). Kinematic data were used to determine total distance and area covered. Peak velocity and movement units (MU), consisting of one acceleration and one negative acceleration throughout a motion [2], were calculated.

One-way ANOVAs were used to determine differences between TD subject's arms and CP subject's arms with Tukey's post-hoc analysis. An alpha level of 0.05 was used to determine statistical significance.

RESULTS

Total distance throughout all of the personal workspace motions was significantly higher for the univolved limb of impaired subjects compared to their impaired limbs and both limbs of typically developing children (p=.0009). There were no significant differences in peak velocity, or MU for dominant and non-dominant arms of TD subjects.

There were no significant differences in peak velocity during the HtH motion (p=.1131). Peak velocity was significantly lower for the uninvolved limbs of CP subjects than their hemiparetic limb, or those of TD subjects during the HtM motion (p=.0112). Peak velocity during the HtB motion was significant (p=.0275) with no differences shown in post-hoc analysis. Peak velocity was significantly higher during the HtP motion for both limbs of the CP limbs compared to TD limbs (p=.0002). Movement units were significantly higher for both limbs of impaired subjects compared to TD subjects for all motions (Fig.1) (HtH (p<.0001), Hand to Mouth (p<.0001), HtB (p<.0001), HtP (p<.0001)).

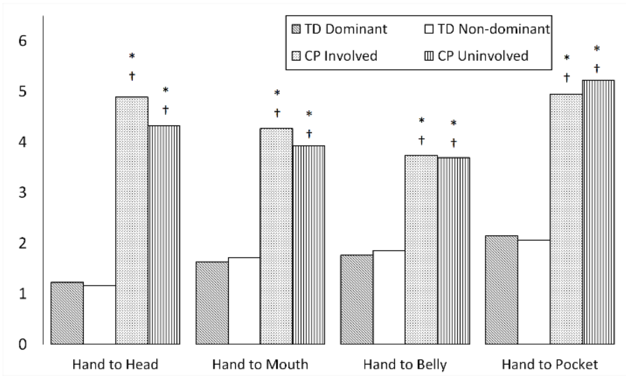

Figure 1. Movement units for each of the inner workspace tasks for typically developing (TD) dominant and non-dominant hands and cerebral palsy (CP) involved and uninvolved limbs. * indicates different from TD dominant limb, † indicates different from TD non-dominant limb.

DISCUSSION

For both limbs, CP subjects showed an increase in MUs over their TD counterparts. This indicates both limbs are unable to perform typically smooth motions and are likely both hindered in motor control, regardless of hemiparetic involvement. Previous studies have shown similar results, as well as improvement in smoothness of motion with treatment [2]. Peak velocity appears to be an inconsistent factor in regards to a personal workspace. Personal workspaces may prove useful in assessing tasks associated with ADLs. Further studies assessing relationships between personal workspaces and ADLs are needed.

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DISCLOSURE STATEMENT

The authors have no conflicts of interest and nothing to disclose.

PODCI Scores within GMFCS Level in Individuals with Cerebral Palsy

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INTRODUCTION: The Pediatric Outcomes Data Collection Instrument (PODCI) is a patient/parent reported outcome measure widely used in children with cerebral palsy (CP). PODCI scores have been compared for patients of different Gross Motor Function Classification System (GMFCS) levels [1]. However, PODCI score variability has not been examined in patients of GMFCS level IV. The purpose of this study is to examine the distribution of PODCI scores within patients with CP GMFCS level I-IV.

CLINICAL SIGNIFICANCE: PODCI scores have not been examined in this manner for individuals that are GMFCS level IV. Utilizing a patient/parent reported outcome measure is useful for providing patient centered care for areas that are meaningful to the family.

METHODS: We retrospectively identified all patients with CP seen for gait analysis at our institution who had been assigned a GMFCS level and whose parent/caregiver had completed the PODCI at their visit. A total of 214 subjects met the eligibility requirements, 35 were GMFCS I, 105 were II, 43 were III, and 31 were IV. Mean age was 11.1, SD 3.7 years (range 4.1-18.9). When patients had more than one visit, their first visit with available data was selected. Patients were grouped based on GMFCS level. One-way ANOVA with pairwise Bonferroni-adjusted post hoc tests was performed to compare the effect of GMFCS level on PODCI scores.

RESULTS: Global, Sports, Transfer, and Upper Extremity scores were different between all GMFCS levels ($p<0.05$) except for upper extremity scores which did not differ significantly between GMFCS levels II and III ($p=0.2$) (Table 1, Figures 1-4). Significant differences were not found for Happiness, Expectations, and Pain scores. Patients in GMFCS level IV had lower Global and physical function (Sports, Transfer, and Upper Extremity) scores compared with all other GMFCS levels. Scores for Happiness, Expectations, and Pain did not differ significantly between GMFCS IV and the other groups although the GMFCS IV group had the highest Expectations and Pain scores and the lowest Happiness scores.

Table 1: PODCI score distribution across GMFCS levels

	GMFCS I N=35	GMFCS II N=105	GMFCS III N=43	GMFCS IV N=31	P
Global	**76.4 (11.9)** ẋ‡Δ	**63.1 (14.7)** Ɨ‡Δ	**51.5 (15.5)** Ɨ ẋΔ	**38.2 (12.6)** Ɨ ẋ‡	**<0.0001**
Happiness	77.7 (21.4)	73.3 (22.4)	73.1 (28.3)	67.6 (32.2)	0.45
Expectations	83.3 (16.9)	82.0 (20.3)	84.8 (19.1)	87.4 (12.2)	0.55
Pain	71.4 (25.8)	70.5 (25.6)	72.7 (24.2)	76.6 (20.6)	0.67
Sports	**66.0 (14.8)** ẋ‡Δ	**46.3 (16.2)** Ɨ‡Δ	**30.1 (15.2)** Ɨ‡Δ	**19.3 (12.3)** Ɨ ẋ‡	**<0.0001**
Transfer	**90.0 (9.7)** ẋ‡Δ	**74.2 (16.3)** Ɨ‡Δ	**50.5 (19.0)** Ɨ ẋΔ	**23.9 (19.4)** Ɨ ẋ‡	**<0.0001**
Upper Extremity	**78.2 (14.4)** ẋ‡Δ	**61.3 (23.7)** ƗΔ	**52.6 (23.6)** ƗΔ	**33.0 (25.0)** Ɨ ẋ‡	**<0.0001**

Scores are reported as mean (SD). Symbols refer to differences between groups, scores in bold represent statistical significance ($p<0.05$) (Ɨ different than I, ẋ different than II, ‡ different than III, Δ different than IV).

DISCUSSION: The utilization of self/parent reported outcome measures allows for more patient/family centered assessments to ensure all of their concerns are being addressed. When these measures are examined alongside functional classification systems, they allow for a more complete clinical picture of the patient, improving treatment recommendations and subsequent outcomes. We found similar results to those in Barnes 2008 study [1], who found that PODCI global, transfers and sports scores were related to GMFCS levels for patients level I-III. However, we included patients GMFCS level IV in our analysis and found that PODCI upper extremity scores were significantly different between level IV and all other GMFCS levels although patients GMFCS levels II and III did not differ from each other in upper extremity scores. Interestingly, while patients in GMFCS level IV clearly had lower functional abilities than patients at lower GMFCS levels, there was much less difference in psychological outcomes like pain perception, happiness, and expectations. This highlights that functional outcomes and psychological outcomes can often be independent from one another. Examining the differences between GMFCS levels for PODCI individual scores allows for health care providers to intervene in areas that are of meaning to these patients and their families.

Figure 1

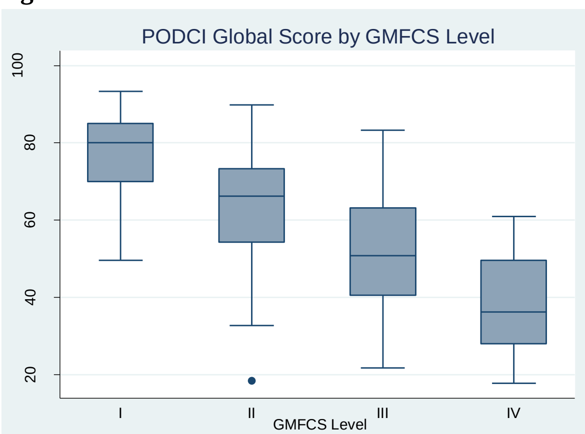


Figure 2

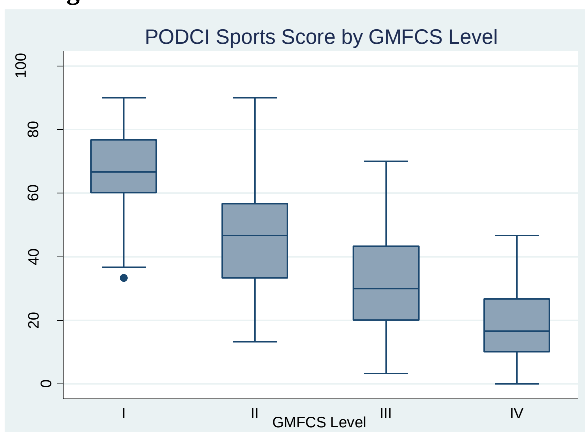


Figure 3

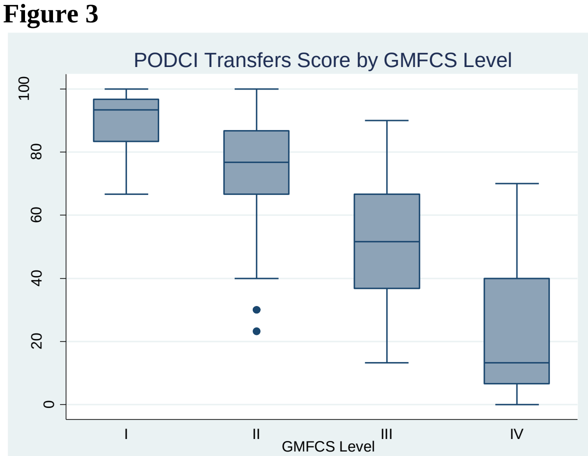
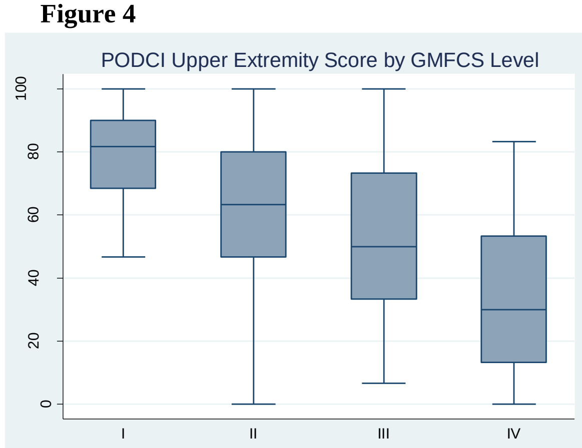


Figure 4



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DISCLOSURES STATEMENT: The authors of this abstract have no disclosures.

GAIT ANALYSIS UTILITY IN FUNCTIONAL GAIT DISORDER

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PATIENT HISTORY

The subject is a 13-year-old female referred for gait analysis to evaluate out-toeing and shuffling gait. She is a previously healthy girl with onset of knee pain 4 years prior, that worsened and progressed over the past 6 months. She is unable to walk comfortably, is having significant pain, and feet turn outward. She is an avid dancer but is now unable to participate. Her only medication is a vitamin D supplement. She had a neurology consult and there was concern for possible neuropathy or dystonia. She underwent EMG nerve conduction studies that were normal. Further diagnostic testing was ordered by orthopedics including hip and knee radiographs, MRI of spine and knees. All were normal. She participated in a course of outpatient PT for 3 months without improvements. A referral to gait lab was made for additional diagnostic evaluation.

CLINICAL DATA

Physical exam was significant for femoral retroversion (outward hip rotation 70°) and outward tibial torsion (thigh foot angle 35°) with weakness in hip musculature and ankle dorsiflexors. She had BMI 14.9, 1%. Functional outcome measures: Gillette Functional Assessment Questionnaire (FAQ): 7; Functional Mobility Scale (FMS): 5,5,5; Pediatric Outcomes Data Collection Instrument (PODCI) scores: Upper extremity: 100, Transfers: 61, Sports: 22, Comfort/pain: 8, Global: 48 Happiness: 25; Bruininks-Oseretsky Short Form administered and she scored in the 2%, well below average. Goal Attainment Scale (GAS): 1. Increased speed and decrease pain. 2. Go back to normal daily activities. 3. Be able to participate in sports and other activities.

MOTION DATA

Gait Analysis: GDI: Left=49.0 Right=44.6 Gait Velocity: 0.34 m/s, 25%. Gait showed increased hip flexion at push-off, decreased peak knee flexion in swing and significant outward hip rotation and foot progression. Kinetics showed reduced power generation at push-off at hip, knee, and ankles.

TREATMENT DECISIONS AND INDICATIONS

Gait analysis revealed an unusual gait pattern and weakness of unknown etiology. Functional gait disorder (FGD), previously known as conversion disorder was suspected. Gait lab recommended psychology consultation and nutrition consult due to very low BMI. A repeat gait assessment was recommended to determine if pattern could be replicated.[1] Although, she did demonstrate true femoral retroversion and outward tibial torsion this was viewed as unlikely etiology of her stiff legged gait and lack of power generation.

OUTCOME

Subject underwent admission at psychiatric hospital for anorexia and FGD. She was placed on Prozac and Ativan. She returned for repeat gait study 8 months after her initial study. The family reported significant improvement in gait and reduction in knee pain. Family rated her listed goals respectively as met, exceeded and greatly exceeded on the Goal Attainment Scale. Functional outcome measures are improved (Table 1) and significant improvement in gait kinematics (Figure 1).

Table 1	Pre-treatment	Post-treatment
Functional Mobility Scale	5,5,5	6,6,6
FAQ: Walking Ability	7/10	10/10
Self-Selected gait velocity	0.34m/s 26% of normal	0.98 m/s 74% of normal
Endurance	½ block	1 mile
Bruininks-Oseretsky	Total = 46 Centile= 2%	Total = 67 Centile=16% 16%
PODCI	Patient Scores	Patient Scores
Upper	100	100
Transfers	61	100
Sports	22	85
Comfort/Pain	8	92
Global	48	94
Happiness	25	45
GDI	L=49 R=44.6	L=91.6 R=85.5

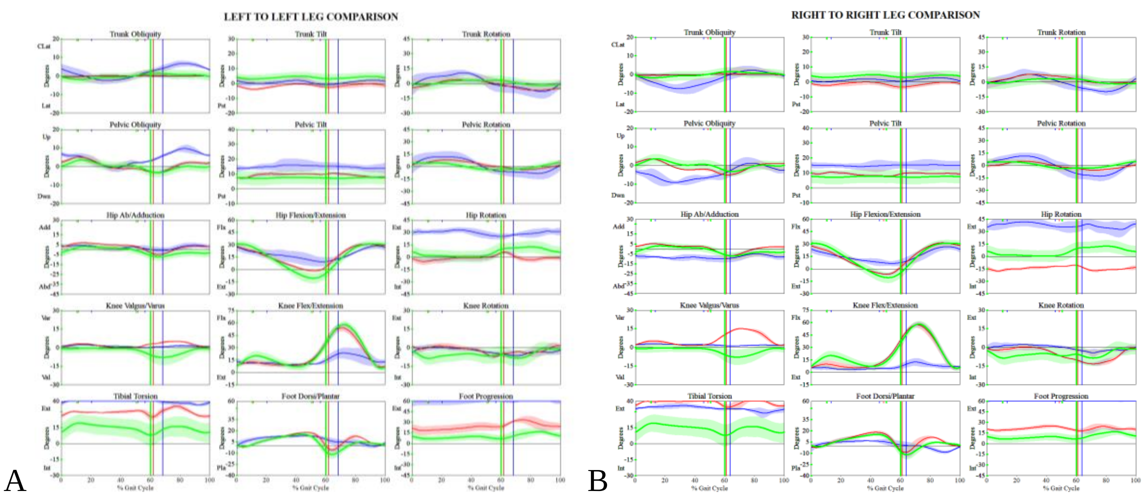

Figure 1. Blue: pre-treatment; Red: post-treatment; Green: controls. A. Left, B. Right Side.

SUMMARY

The atypical gait pattern aided in recommendations for psychological treatment and diagnosis of FGD. FGD has inconsistency and variation in clinical presentation. The disorder is a diagnosis of exclusion, requiring physical causes of symptoms to be ruled out.[2] Inconsistency can occur as dystonic or unusual gait patterns. Previous studies showed there are no single pathognomonic gait patterns. There is unexplained variability and patients shows more marked walking difficulties when being observed.[3]

Gait analysis can document atypical gait patterns and serve as baseline assessment that can be repeated. There was potential her underlying outward tibial torsion and femoral retroversion did cause knee pain as she had abnormal knee moments However, underlying psychological disorder may have distorted her pain perception or caused leading to functional gait disorder. EMG analysis was not utilized but would be recommended in future studies in this patient population. Addition of other measures recommend in the literature included walking with eyes closed, tandem walking, chair sign, pull test, and walking with dual task.[3]

DISCLOSURE STATEMENT: Nothing to disclose.

REFERENCES: 1. *Funct Neurol*. 2012;27(4)., 2. *Neuropsychiatr Dis Treat*. 2005;1(3):205-209. 3. *Neurology*. 2020;94(24):1093-1099.

Real-Time Biofeedback Increases Hip Extension Angle in Individuals After Stroke

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INTRODUCTION

Stroke is the leading cause of serious long-term disability in the United States[1]. Treadmill rehabilitation protocols have shown that real-time visual biofeedback can be used to improve gait mechanics, as it makes individuals post-stroke more aware of how changes to their movement can improve their walking [2]. However, these treadmill-based studies require complex and expensive instrumentation typically limited to research laboratories, making their widespread clinical translation difficult. A recent study found that visual biofeedback during overground walking induces an increase in walking speed in individuals post-stroke [3], demonstrating the potential of translation of visual feedback to overground walking. Hip extension angle is a key biomechanical variable that can be easily measured outside the laboratory using low-cost equipment and has been related to walking speed affecting propulsive forces [4]. Therefore, our aim was to determine the short-term response to hip extension visual biofeedback in individuals post-stroke during unconstrained overground walking. We hypothesized that paretic peak hip extension angle, peak propulsive force, and walking speed will increase from pre-training to post-training. In addition, we hypothesized that a change in walking speed will have a positive relationship with peak hip extension angle.

CLINICAL SIGNIFICANCE

This study creates and tests an innovative visual biofeedback paradigm for individuals post-stroke to increase walking speed. The paradigm requires minimal, inexpensive equipment, can be applied during overground walking, and is completely wearable and portable. The visual biofeedback could provide individuals a simple way to visualize a specific aspect of their walking deficit, enabling them to better understand their own gait patterns and how to change them. Biofeedback-induced increases in walking speed could enhance community ambulation, improve independence in activities of daily living, and increase quality of life of stroke survivors. In addition, portable biofeedback for overground walking has the potential to be modified and target other gait deficits and neuropathologies.

METHODS

Nine individuals post-stroke completed the study protocol (59.09 ± 14.14 years, 5 F, 5 right paresis, 2.36 ± 2.11 years. since stroke). Inclusion criteria were age between 19-80; single-chronic stroke and ability to walk independently or with a cane/walker. Marker positions (Motion Analysis Corp., Rohnert Park, CA) and ground reaction force data from in-ground force platforms (AMTI, Watertown, MA) were collected in separate trials for at least 5 clean foot strikes at 100 and 1000 Hz, respectively, before and after training with the visual biofeedback device. For device training, participants walked for three different six-minute walking sessions. Visual biofeedback consisted of a display attached to eyeglasses that showed a horizontal bar indicating the user's current hip extension angle along with a second horizontal bar symbolizing the target hip extension angle that participants were trying to reach during training. The biofeedback was intermittent, with one minute on and one minute off, to limit dependence on the feedback and to promote motor learning [2]. Wilcoxon signed rank tests were performed for peak hip extension, peak propulsion, and walking speed between pre-training and post-training, to compare for changes in gait mechanics before and after all training sessions. An alpha value of $\alpha \leq 0.05$ was used to determine statistical significance.

RESULTS

Peak hip extension angle was significantly higher after training than before training (Z =-2.19, p=0.03). In addition, peak propulsion was significantly higher after training than before training (Z=-2.31, p=0.02). Walking speed was not significantly higher after training than before training (Z=-1.48, p=0.14) (Figure 1). The change in peak hip extension angle from before to after training was not related to the change in walking speed from before to after training ($r_s(7)$ =0.38, p=0.31).

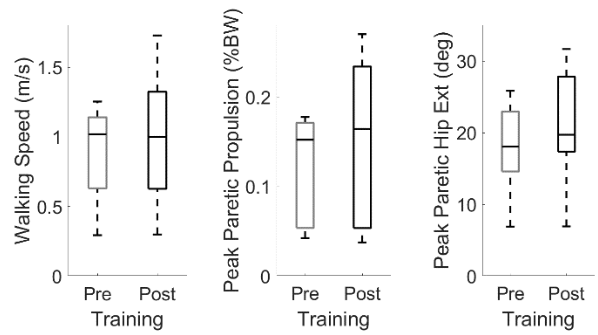

Figure 1. Median walking speed, peak paretic propulsion and peak paretic hip extension angle before and after visual biofeedback training.

DISCUSSION

Our aim was to determine the short-term response to paretic hip extension visual biofeedback during overground walking. We found that 8 of 9 individuals were able to significantly increase their hip extension angle and 7 of 9 individuals increased their peak paretic propulsion and walking speed after training. Interestingly, there was no relationship between the change in paretic hip extension angle and change in walking speed, suggesting that there are various strategies to improve hip extension angle, and some may not be directly related to walking speed. For example, one subject concentrated more on trying to replicate the walking pattern from the training, which caused a reduction in their walking speed. There is also a possibility that the couple individuals that did not increase hip extension, propulsion or walking speed after the training may have been influenced by fatigue. Therefore, a certain cardiovascular fitness may be needed to complete the 18 minutes of training and have the chance to benefit. Future studies can include more participants and investigate what participant characteristics may enable someone to benefit the most from the training.

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THE GAIT OUTCOMES ASSESSMENT LIST (GOAL) QUESTIONNAIRE DEMONSTRATES RESPONSIVENESS TO CHANGE POST SELECTIVE DORSAL RHIZOTOMY IN CHILDREN WITH CEREBRAL PALSY

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INTRODUCTION: The Gait Outcomes Assessment List questionnaire (GOAL) is a valid, reliable parent-reported outcome assessment of family priorities and functional mobility for ambulatory children with cerebral palsy (CP).[1-3] It distinguishes both by Gross Motor Function Classification System (GMFCS) level,[2-4] but responsiveness to change after intervention has not been previously reported. The purpose of this study is to evaluate the responsiveness of the GOAL to change after selective dorsal rhizotomy (SDR) in children with CP.

CLINICAL SIGNIFICANCE: The ability to detect change over time is an important step to determine a 'minimal clinically important difference' in GOAL domain scores after intervention. As the relationship between family priorities, improved function, and objective gait-changes post intervention become more clear, shared decision-making will be enhanced.

METHODS: Responses from the GOAL questionnaire (parent version 5.0) were analyzed for children with a diagnosis of CP (GMFCS I-III) who underwent SDR for treatment of spasticity. The GOAL was completed as part of 3-D gait analysis pre- and one year post-SDR. A repeated measures t-test was used to evaluate changes in pre- and post-SDR domain and total GOAL scores. Items indicated as at least 'difficult' to perform and 'very important' to improve were considered priorities, and priority proportions were also determined pre- and post SDR. McNemar's test was used to evaluate statistical changes in proportions.

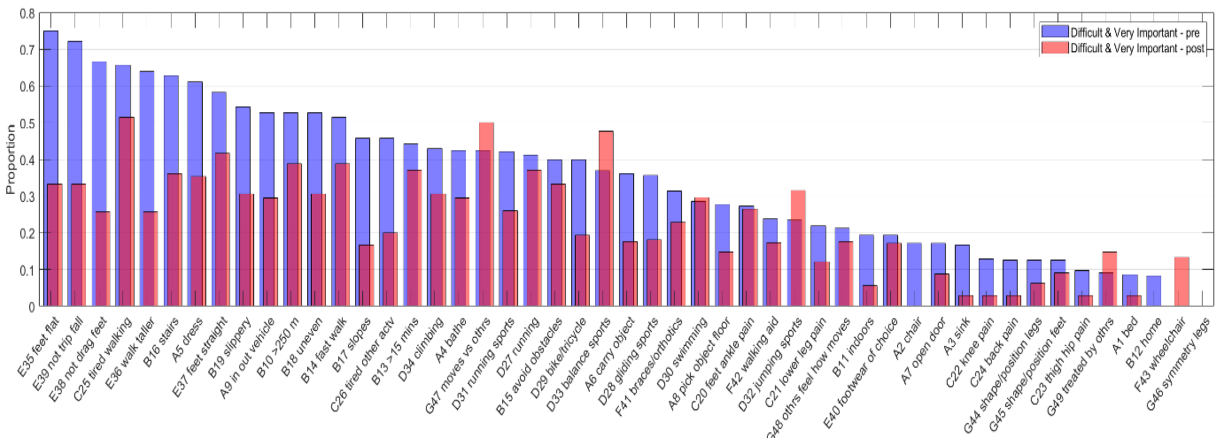
RESULTS: Thirty-six children met criteria for inclusion (23 male; pre-op age 5.6±1.1 years; time from SDR 1±0.1 years; GMFCS I:5; GMFCS II:16, GMFCS III: 15). 83% of the questionnaires were completed by the same parent pre- and post-SDR. The total GOAL score and 4 of 5 functional domain scores (Activities of Daily Living & Independence, Gait Function & Mobility, Pain Discomfort & Fatigue, and Gait Pattern & Appearance) significantly improved pre- to post-SDR (Table). The domains related to happiness (Use of Brace & Mobility Aids and Body Image & Self Esteem) showed no statistical change (Table). Four of five top family priorities, all in the Gait Pattern and Appearance domain ('walk with feet flat', 'walk without tripping/falling', 'walk without dragging feet', and 'walk taller') also significantly decreased 38-40%. 'Feeling tired while walking' did not change as a priority and more than 50% of the families continued to rank it as a top priority goal (Figure). New top priority goals post-SDR included 'the way child moves compared to others, 'participating in sports that require balance', 'walking with feet straight', 'walking faster', and 'walking >250 meters'.

DISCUSSION: The GOAL questionnaire was able to detect significant changes in functional mobility and family priorities post-SDR, suggesting that many GOAL items became less difficult and or less important to improve. The influence, if any, of maturation on the changes is not clear. Regardless, responsiveness to change was demonstrated. This study provides an important baseline to establish a 'minimal clinically important difference' for change in GOAL domain scores after intervention.

Table:

GOAL scores Before/After Selective Dorsal Rhizotomy			
Domain	**Pre-SDR**	**Post-SDR**	**p<.01**
A Activities of Daily Living & Independence	51.2	62.9	*
B Gait Function & Mobility	45.4	54.0	*
C Pain Discomfort & Fatigue	72.5	81.9	*
D Physical Activities Sports & Recreation	22.8	27.5	
E Gait Pattern & Appearance	29.3	50.0	*
F Use of Braces & Mobility Aids	48.7	46.9	
G Body Image & Self Esteem	52.9	51.3	
TOTAL Score	**47.4**	**55.7**	*

Figure



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DISCLOSURE STATEMENT

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ASYMMETRIC MUSCLE ACTIVITY IN PATIENTS WITH PATELLOFEMORAL INSTABILITY

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INTRODUCTION

Patellofemoral instability (PFI) is the most common knee condition during the growth and development years among the pediatric population [1]. Following a dislocation event, up to 40% of children experience recurrent dislocations which ultimately require surgical reconstruction to stabilize the patella. Physical therapy is an important component of non-operative management and post-operative rehabilitation with the aim of restoring joint motion and muscle strength to prevent future dislocation events. However, therapy may often be based upon theoretical best protocols, without robust data on actual muscle weakness or imbalances in the PFI population. At present, muscle activity patterns during common activities of daily living remain relatively unknown in pediatric PFI patients. The purpose of this study was to identify pre-treatment muscle activity patterns during low-impact tasks in patients diagnosed with PFI. We hypothesized that muscle activity would significantly differ between the affected and unaffected limbs; and secondly, that muscle activity would differ between patients with low- and high-recurrence of instability events.

CLINICAL SIGNIFICANCE

Muscle weakness, particularly in the quadriceps, and the resulting compensatory strategies can persist for years following instability events. Identification of muscle activation patterns in this vulnerable patient population will offer physicians and therapists valuable information to support treatment decisions and rehabilitation efforts.

METHODS

Patients were recruited from a single sports medicine clinic, and tested following a recent patella dislocation prior to treatment. Potential participants were excluded if knee effusion was noted. The most recent dislocation event defined the affected side, and total number of dislocations categorized patients into low (≤3) and high (>3) recurrence groups. Muscle activity was measured via electromyography (EMG) sensors placed bilaterally on the rectus femoris (RF), vastus medialis obliquus (VMO), biceps femoris (BF), tibialis anterior (TA), and gastrocnemius (GAS). Subjects performed three trials of overground walking and a step down tap task (SDT). The SDT required subjects to stand on top of a 21cm bench, extend one foot out to the front, lightly tapping the heel of the extended leg onto the floor while keeping their weight on the stance limb, and then return to the starting position. EMG signals were rectified and filtered to compute a linear envelope, and mean and maximum values were calculated from a representative trial per task. Additionally, to compare between groups, mean values were normalized by the maximum value recorded per task. Paired and independent samples t-tests were performed to identify side-to-side and between-group differences, respectively (α=0.05).

RESULTS

Fourteen patients (7 males, 14.3±1.7 years, 8 low-recurrence, 10 pre-operative) were tested 37.5±34.5 days post-injury. Across all patients, the affected limb exhibited significantly reduced RF activity during gait (affected: 8.4±3.5μV, unaffected: 13.4±8.8μV, p=0.04) and VMO activity during the SDT (affected: 90.4±53.9μV, unaffected: 132.0±69.0μV, p<0.01) as shown in Table 1. Furthermore, the high-recurrence group exhibited increased RF activity on the unaffected limb during gait compared to the low-recurrence group (Low: 31.0±7.6 %max, High: 48.4±17.6 %max, p=0.03) (Figure 1).

Table 1: Muscle activity (μV) for the Affected and Unaffected limbs; Max (SD). * α<0.05

	Affected	**Unaffected**	***p*-value**
Walking			
RF	8.4 (3.5)	13.4 (8.8)	**0.04***
VMO	20.9 (18.4)	24.1 (22.0)	0.71
BF	22.9 (12.4)	22.3 (10.1)	0.89
TA	66.1 (24.8)	83.2 (47.0)	0.19
GAS	64.7 (32.3)	70.7 (36.1)	0.57
SDT			
RF	47.5 (19.6)	58.2 (25.7)	0.31
VMO	90.4 (53.9)	132.0 (69.0)	**<0.01***
BF	28.2 (17.8)	24.3 (9.6)	0.47
TA	83.0 (40.5)	85.8 (46.8)	0.74
GAS	40.9 (25.0)	43.2 (32.8)	0.86

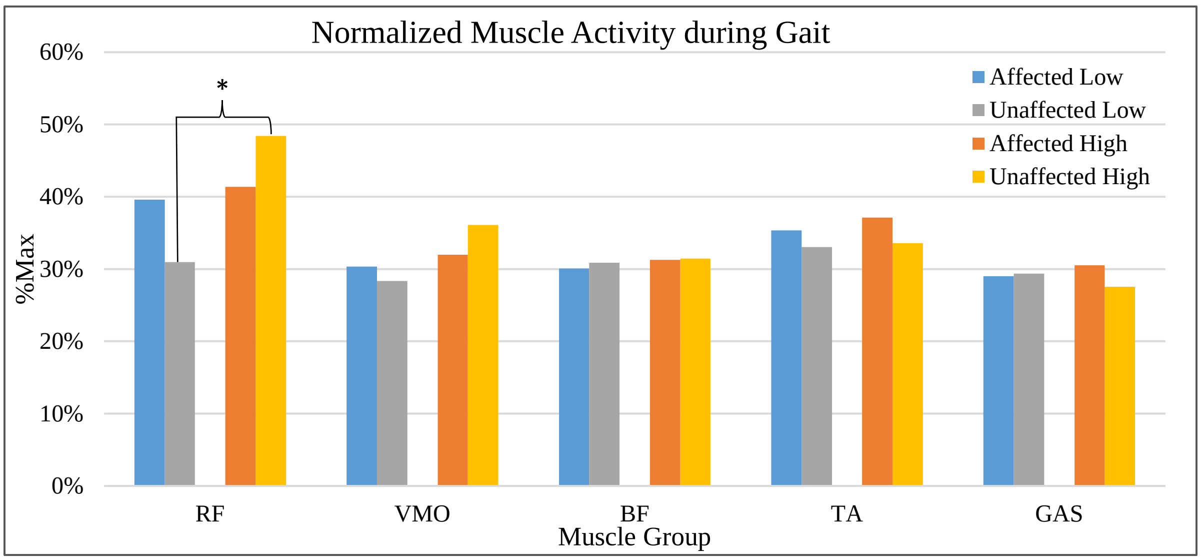

Figure 1: Normalized muscle activity for low recurrence (Low) and High recurrence (High) groups for the affected and unaffected limbs during gait. (* denotes significant finding)

DISCUSSION

Reduced RF and VMO activity on the affected limb was observed during walking and a step down tap task compared to the unaffected limb. Additionally, patients who have experienced a higher recurrence of dislocation events may develop compensatory strategies that rely more on the quadriceps of the unaffected limb during gait to avoid potential re-injury. Further kinematic analysis is needed to better understand the observed asymmetries.

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DISCLOSURE STATEMENT Co-author Tulchin-Francis is a GCMAS executive board member. No other conflicts of interests to disclose.

COMBINED AUDIOVISUAL AND HAPTIC BIOFEEDBACK ALTERS THE RATE AND MAGNITUDE OF PLANTARFLEXOR ADAPTATION IN CEREBRAL PALSY

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INTRODUCTION

Biofeedback is a promising non-invasive strategy to enhance gait training in cerebral palsy (CP). Commonly, biofeedback systems are designed to provide extrinsic feedback on movement errors using audio or visual cues; these systems have been previously used in CP to improve both kinematics and muscle activity [1]. More recently, haptic feedback (HF) systems, such as exoskeletons, have become popular as a means of promoting reliable intrinsic feedback, which may be especially important in CP where sensory processing is impaired [2]. To this end, recent work has demonstrated that training with an ankle exoskeleton can improve motor control during walking in children with CP [3]. While the success of these modalities highlights the efficacy of using biofeedback for rehabilitation, combining modalities may further amplify outcomes. Prior research has demonstrated that robust error recognition drives adaptation, suggesting that biofeedback systems that present multisensory cues may increase the rate and magnitude of response [4]. However, it is largely unknown how individuals with CP adapt gait using biofeedback and how the choice of modality may influence this ability.

The aim of this study was to compare how individuals with CP adapt plantarflexor activity while walking with audiovisual feedback (AVF) and HF, independently and in combination. Secondarily, we evaluated changes in antagonist co-contraction to understand if walking with biofeedback promoted undesirable compensation. We hypothesized that combining AVF and HF would promote greater error recognition than either modality alone, increasing both the rate and magnitude of adaptation.

CLINICAL SIGNIFICANCE

This study demonstrates that the modality of biofeedback influences the extent to which individuals with CP modulate plantarflexor activity, which may inform future system design.

METHODS

Eight individuals with diplegic CP (14.3±2.0 yrs; GMFCS I-III) walked at a self-selected speed (0.81 ± 0.10 m/s) using three biofeedback modalities: (1) AVF, (2) HF, and (3) AVF + HF. Trials were performed in a randomized order and included baseline (1 min.) feedback (6 min.) and washout (1 min) phases.

Both AVF and HF targeted soleus activity on the more affected limb. HF was administered using a lightweight, battery-powered ankle exoskeleton that has been previously evaluated in CP [3]. This system imparts a *resistive* ankle moment during push-off proportional to the estimated biological ankle moment. The AVF system was custom designed to display a real-time trajectory of soleus activity alongside a target score and play a tone each time the score was reached. To balance motivation and challenge, the target score was programmatically adjusted to maintain participant success rate between 50-75%. During the AVF-only condition, participants wore the exoskeleton in the zero-torque mode to reduce any device weight effects.

Bilateral electromyography (EMG) data were collected from the soleus and tibialis anterior for all trials (1000 Hz; Noraxon, Scottsdale, AZ). EMG data were high pass filtered (40 Hz), rectified, and low pass filtered (10 Hz). Data were then normalized to the mean activity from the first baseline phase tested for each participant. Stance phase soleus activity for the targeted limb was quantified for five-stride windows over the length of the feedback phase to capture transient responses to each modality. Further, ankle co-contraction for each window was measured using the co-contract index (CCI) [5]. Wilcoxon signed-rank tests with a Holm-Šídák correction for multiple comparisons (n = 3) were used to identify deviations in soleus activity and CCI from baseline for each five-stride window across modalities (α = 0.05).

RESULTS

The rate and magnitude of soleus adaptation differed across biofeedback modalities. Both AVF and AVF + HF resulted in a sustained increase in soleus activity during the feedback phase ($p < 0.046$). However, AVF + HF had the largest overall increase in soleus activity (median [IQR]: 56.2% [15.2,66.8]) compared to AVF (40.6% [21.7,60.9]) and HF (11.9% [7.1,33.2]). Individuals walking with AVF + HF also adapted more quickly than with the other modalities, significantly increasing soleus activity within the first five strides of exposure. Interestingly, the apparent advantage of AVF + HF was not maintained, returning to magnitudes similar to AVF after just twenty strides. Adaptation to HF was also transient, returning to baseline values after twenty-five strides ($p > 0.078$). No modality demonstrated a significant washout effect ($p > 0.29$), suggesting that gains were context-specific. Further, CCI did not change for any modality, indicating that individuals modulated soleus activity without increasing co-contraction ($p > 0.38$).

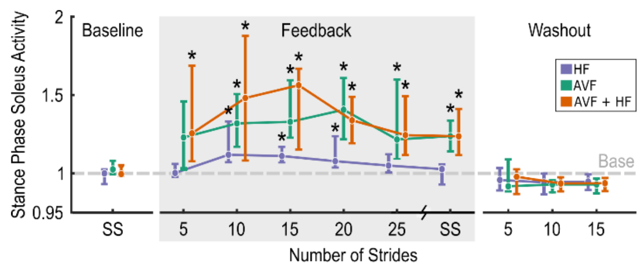

Figure: Median [IQR] soleus activity in the targeted limb for 5-stride windows and steady state (SS) walking. * denotes significant difference from baseline.

DISCUSSION

Both AVF and AVF + HF consistently increased soleus activity in a single session, but individuals adapted more quickly to AVF + HF, demonstrating the advantage of pairing modalities to rapidly improve error recognition and subsequent correction. Unexpectedly, response to HF diminished over time which could suggest that individuals were differentially prioritizing AVF and HF input signals. Alternatively, because HF created more challenging walking conditions, these findings could also suggest that additional acclimation is necessary to promote consistent device engagement. Further work is needed to understand if response to each modality changes with additional training and what factors may underlie existing inter-participant variability.

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DISCLOSURE STATEMENT

Zachary F Lerner is a co-founder with shareholder interest in a start-up company seeking to commercialize the ankle exoskeleton in this study. There are no other competing interests.

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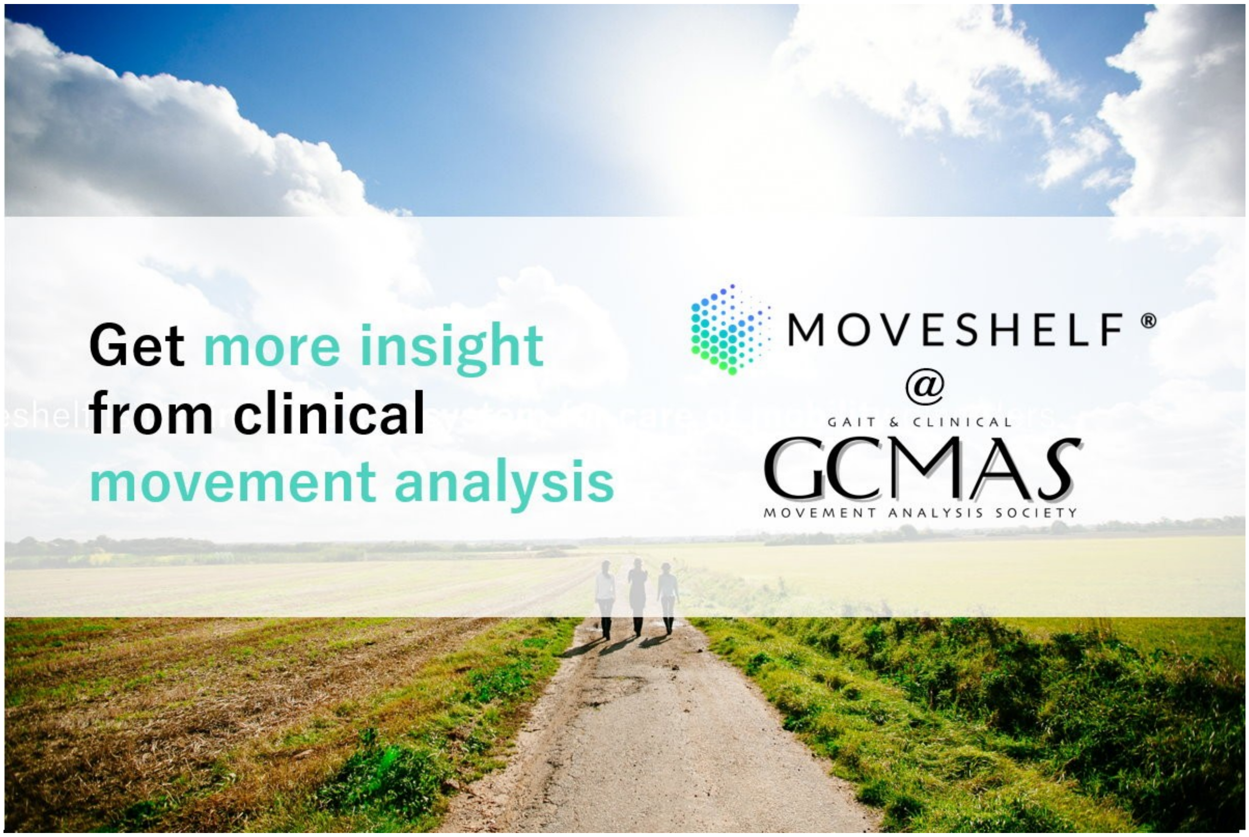

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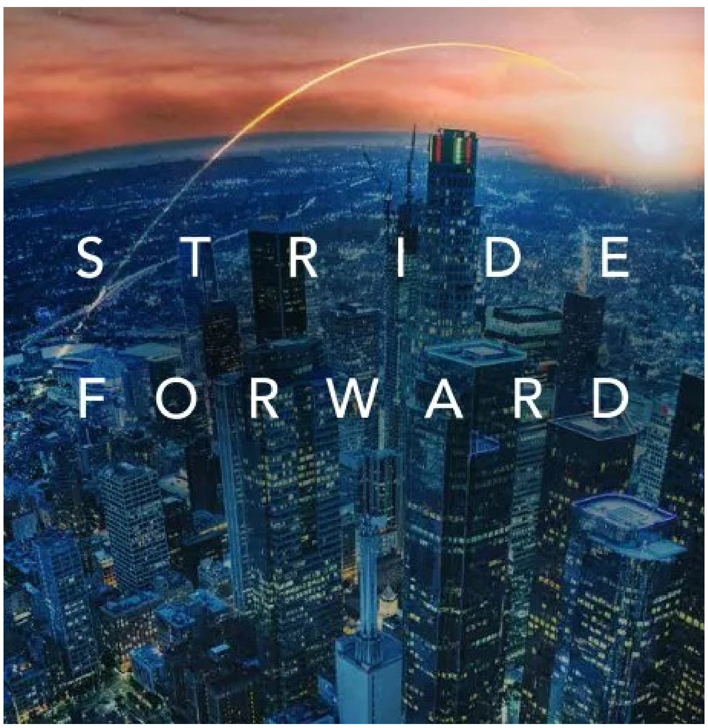

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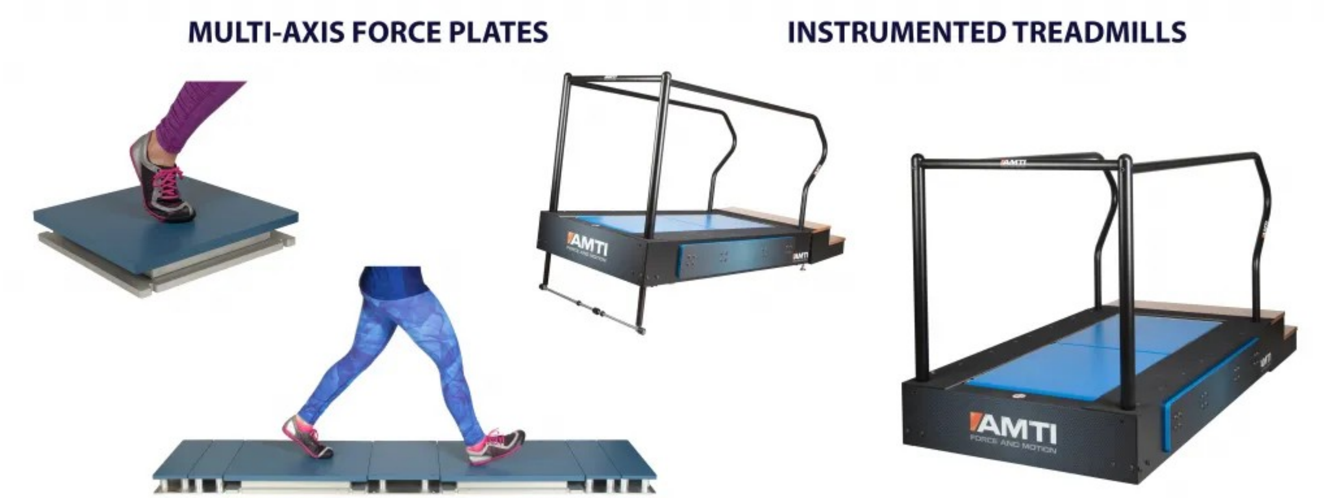

In Memory of Sara J Mulroy

Contributed by Walter Weiss during GCMAS2022

It is with profound sadness that we share of the passing of our colleague and friend, Sara J. Mulroy, PT, PhD, on May 3, 2022. Dr. Mulroy's contributions to Rancho, our patients and our research legacy cannot be overstated. As a valued member of the Physical Therapy Department at Rancho Los Amigos National Rehabilitation Center since 1986, Dr. Mulroy dedicated her life to the Rancho way: always trying to find ways through her years of clinical service and research to help improve the lives of those around her. She joined the Rancho Physical Therapy Department in 1986 when she began PhD training under Dr. Jacquelin Perry, then followed Dr. Perry as the Director of the Pathokinesiology Lab in 1996. Over the years, Dr. Mulroy's passion for helping those with disability has been evident through her personal involvement and oversight of much of Rancho's paradigm-shifting physical therapy research aimed at improving function, participation and quality of life for those living with post-polio syndrome, stroke, amputations, and spinal cord injury. Most recently, Dr. Mulroy was the driving force behind multiple nationally funded research projects, including as the Director of Rancho's nationally recognized Spinal Cord Injury Model System. Dr. Mulroy was a guiding light in the area of physical rehabilitation, gait analysis and the recovery of walking after injury, and internationally recognized for her expertise and efforts to improve the lives of those with disability. Among the many honors she has received are the MarianWilliams Award for Research in Physical Therapy from the American Physical Therapy Association and the Joseph F. Dowling Distinguished Service Award from Rancho. Most importantly, Dr. Mulroy has been an exceptional friend, colleague and mentor to all those around her. Her light, dedication to helping others, and optimistic spirit will be greatly missed by all of those that knew her. Dr. Mulroy is survived by her husband Kevin, son Kieran, and brother Tom and his family.

In Memory of Lori A Karol

Written by Dan Sucato, Chief of Staff at Scottish Rite for Children

With additional contributions by Kirsten Tulchin-Frances, President, GCMAS

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Lori A. Karol, MD, professor of Orthopaedic Surgery, University of Colorado School of Medicine and Chief of Pediatric Orthopaedics at Children's Hospital of Colorado; and former Professor of Orthopaedic Surgery, Department of Texas Southwestern Medical School, Assistant Chief of Staff, Texas Scottish Rite Hospital, died Saturday, February 26. With Dr. Karol's passing, we have lost an incredible person, physician, surgeon, and leader whose impact will forever be felt.

Dr. Karol, a native of Detroit, Michigan, completed the integrated premedical/medical program at the University of Michigan—a 6-year program combining undergraduate and medical school studies. She remained a die-hard enthusiastic Michigan fan, always ready to showcase the Maize and Blue wherever and whenever possible. She completed her orthopaedic residency program at Wayne State where she distinguished herself in becoming the chief administrative resident and winning the AOA-Zimmer Resident Research Award given at the American Orthopaedic Association Meeting. Dr. Karol moved on to Dallas for her Fellowship in Pediatric Orthopaedics and Scoliosis at Texas Scottish Rite Hospital. Following fellowship, she was recruited to California and was an Assistant Professor at the University of California, Davis, for three years before being recruited to return to TSRH by Dr. Tony Herring, then Chief of Staff, where she practiced for the next 27 years. Dr. Herring writes, "She was a trailblazer in the finest sense of the word. In her straightforward approach to every task, big or small, she did the work and always got the job done without fanfare or glory."

Dr. Karol was the consummate academic orthopaedic surgeon. She was an exceptionally skilled clinician, a brilliant researcher, and a natural educator.

Dr. Karol started her career at Scottish Rite as a general pediatric orthopaedic surgeon and finished as one. She managed everything from scoliosis to clubfeet to hip dysplasia to limb length discrepancy and performed both nonoperative and operative procedures for all of these conditions. More than her exquisite clinical skills in managing both the straightforward and complex conditions was her ability to "treat a family." Dr. Karol had a gift connecting with patients and their families that was unparalleled and resulted in the highest patient satisfaction scores each and every month. Her clinical practice positively impacted thousands of children with some of the most challenging conditions, and she guided them through with the skill and humanity few can provide, ultimately leading to a better life for each of them.

Dr. Karol advanced her academic standing at the University of Texas Southwestern where she achieved full professorship in the Department of Orthopaedic Surgery in a short 12-year period. She was a prolific researcher— publishing over 100 peer-reviewed papers and presenting greater than 200 podium presentations at national and international meetings. The large volume of scientific work is only surpassed by the quality of the work which has influenced the world of medicine and led to improvements in the care of children worldwide. Her seminal works on pulmonary function in early onset scoliosis, gait and function following clubfoot treatment, optimal treatment of hips in children with cerebral palsy, orthotic management of scoliosis, and energy expenditure for children with amputations have changed the field of orthopaedics for the better. Her work has been recognized by the Pediatric Orthopaedic Society of North America (POSNA), winning the Arthur H. Huene Award (2001) and the Best Paper Award (2015); by the International Congress on Early Onset Scoliosis (ICEOS), winning the Behrooz Akbarnia Best Paper Award (2017); and by the Limb Lengthening and Reconstruction Society (LLRS), winning Best Paper Award (2018).

During her tenure at Scottish Rite, Dr Karol was the medical director of the Movement Science Laboratory for 25 years. In addition to reviewing cases for all clinical patients in the lab, she led many of the lab's research projects evaluating function in children with clubfoot, scoliosis, cerebral palsy, limb deficiency and other conditions. She was a mentor to the staff in the 'Gait Lab', and trained numerous fellows and residents in gait analysis. Dr Karol was a long-standing member of the Gait and Clinical Movement Analysis Society.

In the midst of all of her clinical and academic work, she remained a natural and exceptionally effective educator to medical students, residents, fellows, and all those around her. Whether she was in the clinic, the operating room, the hallway, or at the podium, she was always passing on pearls of wisdom that we all remember and use in our daily lives as orthopaedic surgeons. She could remember diagnoses and patients when cases came up to pass on key points for upcoming challenging treatment decisions for colleagues and former fellows. She was a sought after speaker, having been invited to 41 visiting professorships at leading orthopaedic centers throughout the world where she spoke on a wide array of topics.

Leadership came naturally to Dr. Karol whether it was within her own institution or on the national and international scenes. At Scottish Rite, she was the medical director of the Movement Science Laboratory (a position recently named in her honor), was an Assistant Chief of Staff, and became the first Chief Quality Officer in the hospital's history. The "firsts" continued as she was the first female President of POSNA in 2016 following a long history of service to the society as the chair of several committees and councils. She was honored with the Distinguished Achievement Award in 2021 and was recently inducted into the POSNA Hall of Fame. She was also the first female elected President of the Scoliosis Research Society—a commitment she could not fulfill due to her recent illness. Her leadership skills have always been recognized, and she was recruited several times to become "chief" at children's hospitals. She finally took that step in 2020 when she was recruited to become Chief of Orthopaedics at Colorado Children's Hospital.

Above all of the professional achievements, Dr. Karol was an amazing person—elegant and sophisticated, who by her actions, encouraged all around her to be their best self and who invoked a sense of humanity in all of us. Dr. Karol was a great human being who cherished life, loved Jimmy Buffet, was a voracious reader and a master baker, enjoyed gardening, yoga, and a nice glass of wine. She always took the time to get to know you and your family. She was a great conversationalist with a witty sense of humor and she had an infectious personality. She was an inspiration to everyone, including women surgeons and physicians, by her example of leadership and doing it the right way, while balancing her busy professional and personal life.

Finally, and most importantly, Dr. Karol always placed her greatest importance on her family as a devoted wife, seemingly always with Bob, and most impressively, a wonderful mother to three daughters who have achieved so much in their young careers in accounting, nursing, and law. We say goodbye to a colleague, a leader, an icon, a mentor, and a dear friend who leaves a large void in our lives and in this profession but who we will always be remembered for making us a better person and physician.

We will miss you Lori.