Abstracts of the 25th Annual Meeting

of the

Gait & Clinical Movement Analysis Society

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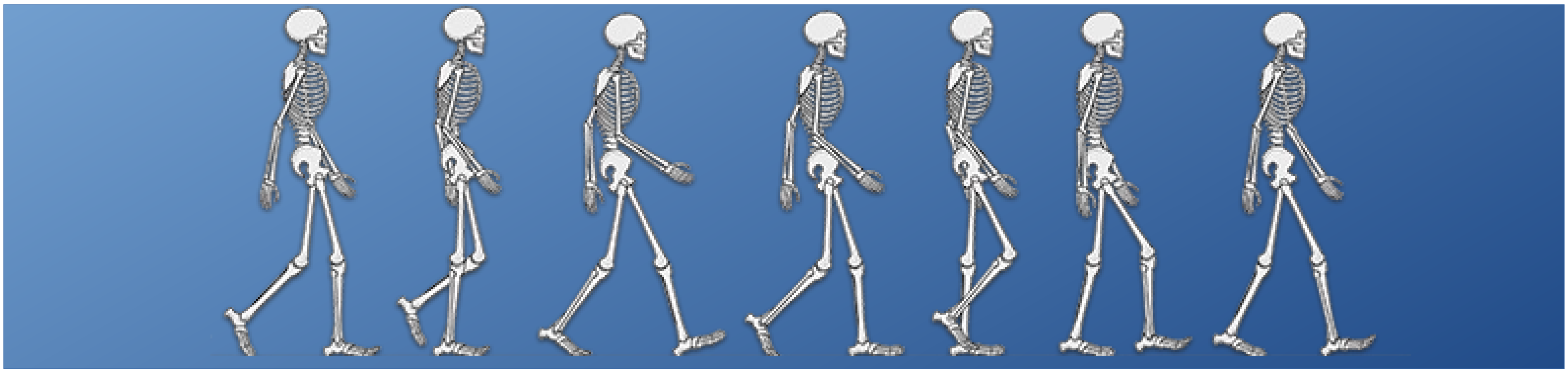

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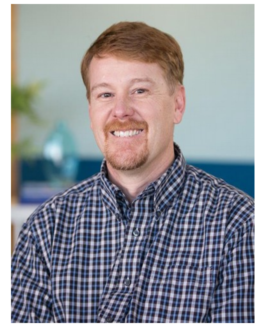
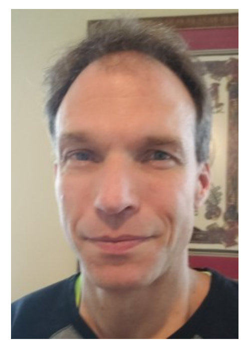


I again send my and the board's apologies for canceling the annual meeting in West Chester this year. All refunds are being processed and you should receive them soon. We will plan to have all abstracts on the GCMAS website in the near future and access will be emailed out to the society members. We have decided to do this so that everyone's exceptional work can be reviewed to continue our learning and scientific sharing within the society. We also feel it is important for all to be able to receive academic credits for the time and work that has been invested in these projects. We plan to have the 2021 in West Chester and details and dates will be released soon. '

Thank you to all and stay healthy!

Jason Rhodes, GCMAS President

Dear GCMAS members and supporters,



Although we are not having the conference this year due to the unprecedented COVID-19 lockdown, we have compiled the abstracts into a program so that you can desseminate your hard work to our community. Abstracts submitted for review as posters or podium papers were reviewed by two or more reviewers and scored by each on a scale of 1-5 based on scientific merit, originality, clarity, and presentation. Chris Church, our program chair, then organized the submissions based on topic into sessions. What is presented here mirrors what would have been presented at the conference, although with slightly greater detail than is typical in the area of posters. Accepted podium papers are presented first by session topic in alphabetical order. Next come the tutorial descriptions. Finally are the posters, also arranged by topic, although at the conference these would have been simply presented over two days. I hope you find this abstract booklet easy to navigate. Care has been taken to present a table of contents, and like all PDF files, it is searchable by text. We have also, for the first time, obtained a DOI for this document so that it may be cited by you as needed.

I look forward to seeing you at GCMAS next year, and thank you for all of your effort and support at this time.

Tim Niiler, GCMAS2020 Conference Chair

Podium-Balance

Podium-Cerebral Palsy

Podium-NeuroMuscular

Podium-Quality Assurance

Podium-Sports

Thematic Poster-Sports

Podium-Upper Extremity

Tutorial

Poster-Case Studies

Poster-Foot Interventions and Biomechanics

Poster-Novel Measurement Techniques

Poster-Orthopedics

Poster-Sports

Poster-Upper Extremity and Trunk

Balance Effort, Cone of Economy, and Dynamic Compensatory Mechanisms in Common Degenerative Spinal Pathologies

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INTRODUCTION

Changes in balance are common in individuals with spinal disorders and may cause falls. Aging, vestibular deficits, poor vision, spinal disorders that cause changes in posture, and a weak core/legs all predispose an individual to postural instability.[1] When one's balance and function are detrimentally altered, it can have a profound effect on their quality of life and activities of daily living.[2] The spinal cord, particularly the dorsal columns, provides integral sensory feedback.[3] The dorsal columns relay positional, two-point discrimination, and vibration as well as play an important role in maintaining postural stability while conveying sensory information, such as deep sensations, to the lower limbs.[3] When the dorsal columns of the spinal cord are compressed, the functions of vibration sense, deep sensibility, and juxtaposition/proprioception are compromised.[3] Recently, a method that quantifies an individual's cone of economy (CoE), first proposed by Prof. Dubossett, using 3D video kinematic data was developed (Figure 1).[4] This method quantifies CoE dimensions by measuring the range of sway (RoS) and balance effort for each patient. Dynamic compensatory mechanisms can be described as emergent neural control processes and are best differentiated by what the central nervous system (CNS) is attempting to control.[5] Although ankle and hip excursion in standing have been identified as common compensatory mechanisms, some studies have described how one strategy or a combination of strategies may be preferred over another in different situations. Based on our knowledge, no study has presented balance effort, CoE quantitative measurements, and dynamic compensatory mechanisms in common degenerative spinal disorders patients. Therefore, the purpose of this study was to discern differences in balance effort values between common degenerative spinal pathologies and a healthy control group.

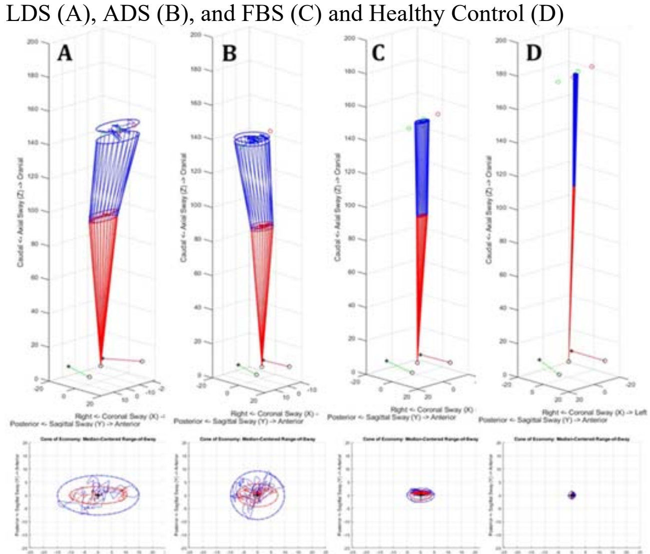
CLINICAL SIGNIFICANCE

Patients with degenerative spinal pathologies exhibit markedly diminished balance and compensatory mechanisms as indicated by increased sway and larger joint excursion on a Romberg test and larger CoE.

METHODS

This study was a retrospective cross-sectional study of symptomatic adult patients and healthy volunteers who received functional evaluations at our institution between 2016 and 2019. Three-hundred and forty patients with one of six degenerative spinal pathologies: 84 cervical spondylotic myelopathy (CSM), 96 adult degenerative scoliosis (ADS), 28 sacroiliac joint dysfunction (SIJD), 58 degenerative lumbar spondylolisthesis (DLS), 46 single-level lumbar degeneration (LD), and 28 failed back syndrome (FBS). Forty healthy control subjects were recruited from the general population. Each patient was tested approximately one week before surgery. Each patient fitted with a full-body external reflective marker set (Vicon, Oxford, UK). Each patient performed a functional balance test. Patients were asked to stand erect with feet together and eyes open in their self-perceived balanced and natural position for a full minute while measures of sway were recorded. This was similar to the Romberg's test. Sway calculation for the Romberg's test was based on previously published balance work (Figure 1).[4]

Figure 1. Representation of Cone of Economy and Range of Sway LDS (A), ADS (B), and FBS (C) and Healthy Control (D)



Three-dimensional CoE boundaries were determined during the functional balance tests. A custom algorithm was used to determine sagittal, coronal, and vertical RoS for the CoM and the head ((Figure 1). To evaluate dynamic compensatory mechanisms, a balance control strategy analysis was developed based on previous methods developed by our lab. Joint angles for the ankle, knee, hip, pelvis, and trunk were calculated at anterior and posterior peak sways. This data was used to categorize dynamic compensatory mechanisms for each patient. Multiple one-way ANOVAs were used to test for significant changes between the degenerative spinal pathologies cohorts to the healthy group. Statistical analyses were conducted using SPSS, Version 23.0 (IBM, Armonk, NY, USA).

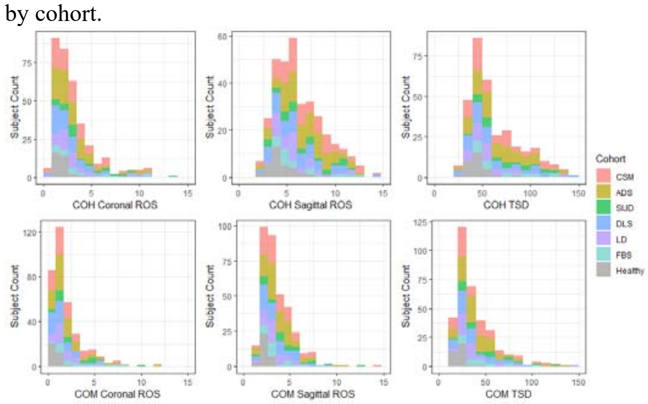
RESULTS

Cone of Economy and Balance Effort: Balance effort found to be significantly larger in degenerative spinal pathologies patients when compared to controls. Head and CoM overall sway ranged from 65.22-92.78 cm ($p<0.004$) and 35.77-53.31 cm ($p<0.001$), respectively in degenerative spinal pathologies patients and in comparison to controls (Head: 44.52 cm, CoM: 22.24 cm). CoE dimensions and RoS found to be greater in degenerative spinal pathologies patients in comparison to controls. Head (6.43-7.79 cm, $p<0.001$) and CoM (3.37-4.39 cm, $p<0.001$) sagittal RoS found to be bigger than controls 4.24 and 2.34 cm, respectively. Head (3.09-5.59 cm, $p<0.046$) and CoM (1.78-3.67 cm, $p<0.022$) coronal RoS found to be bigger than controls 1.81 cm and 0.90 cm, respectively. Head and CoM vertical RoS found to be greater than controls as well (Figure 2). Dynamic Compensatory Mechanisms: Patients with degenerative spinal pathologies found to have more compensatory mechanisms as seen by larger excursion at their trunk, hip, and knee joints. Degenerative spinal pathologies patients presented with greater trunk (1.61-2.98°, $p<0.038$), hip (4.25-5.87°, $p<0.049$), and knee (4.55-6.09°, $p<0.036$) flexion angle during a maximum anterior sway when compared to controls (trunk: 0.95°, hip: 2.97°, and knee: 2.43°). Moreover, degenerative spinal pathologies patients presented with lesser hip extension (0.82-1.36°, $p<0.033$), and more knee extension (-3.02-(4.51)°, $p<0.035$) angle during a maximum posterior sway when compared to controls (hip: -1.28°, and knee: -0.24°).

DISCUSSION

The results of this study indicate that patients with degenerative spinal pathologies exhibit markedly diminished balance (and compensatory mechanisms) as indicated by increased sway on a Romberg test and larger CoE. Balance effort, as measured by overall sway, was found to be approximately double in degenerative spinal pathologies patients in comparison to healthy matched controls. CoE dimensions were found to be significantly greater in common degenerative spinal pathologies patients in comparison to controls. This may be for a number of neurophysiologic reasons which can include: direct inhibition of sensory/proprioceptive feedback (CSM), overcompensation to restore sagittal/coronal balance (ASD), pain mechanisms causing altered posture (SIJD/FBS/LD), and habitual postural changes due to chronicity of symptoms. Conscious and unconscious proprioceptive mechanisms (vestibulospinal tracts, spinocerebellar tracts, dorsal columns) require unimpeded pathways for optimal function. Their alteration, as per our data, reliably demonstrates quantifiable alterations in CoM as well as head sway (overall, sagittal, coronal).

Figure 2. Histogram plots showing the head range and total sway by cohort.



ADS: Adult Degenerative Scoliosis; CSM: Cervical Spondylotic Myelopathy; SIJD: Sacroiliac Joint Dysfunction; DLS: Degenerative Lumbar Spondylolisthesis; LD: Lumbar Degeneration; FBS: Failed Back Syndrom; COH: Center of Head, ROS: Range of Sway.

Degenerative spinal pathologies patients exhibited greater knee, hip and trunk motion which results in higher energy consumption and greater utilization of pelvic and lumbar spine musculature. In comparison to the healthy group, all patients presented with knee and hip-based compensatory mechanisms. In general though, cortical and subcortical mechanisms, through the righting reflex, will sacrifice focal alignment to optimize global balance within the CoE. Each cohort exhibited different compensatory mechanisms however anterior sway for each pathology tended to have a preponderance toward anterior sway which really points toward proprioceptive stimulation of antigravity musculature. Without a vector, falls forward are much more common than backward.

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DISCLOSURE STATEMENT

Drs. Haddas, Kosztowski, Mar, Boah, and Lieberman have nothing to disclose.

OPTIMIZING GAIT OUTCOMES IN PARKINSON'S DISEASE WITH AUDITORY CUES: THE EFFECTS OF SYNCHRONIZATION AND GROOVE

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Introduction

Walking is a naturally rhythmic pattern with a regular and repetitive cycle, much like that of music. The rhythmic nature of gait and music has been capitalized on in neurological rehabilitation to support natural and safe walking patterns among people with impaired gait, in particular in Parkinson's disease (PD). Rhythmic auditory stimuli (RAS), such as metronomes or music, provide temporal information to which a person can entrain their gait. Much work supports that RAS can produce longer, faster strides with increased stability when played at a rate faster than baseline cadence; however, these findings are variable in the literature and consequently pose a barrier to appropriate clinical implementation [1].

Some of this variability may stem from properties of the music and how users interacts with the cue. Perceiving more groove in music, or feeling greater desire to move with music, consistently elicits faster gait with larger strides in healthy adults [2-4]. Furthermore, this appears to depend in part on whether users are actively trying to synchronize with cues or not [4-5]. To date, no work has investigated how groove and synchronization together impact gait outcomes in PD.

Clinical Significance Gait impairments are a debilitating aspect of PD that are associated with increased fall risk, decreased quality of life, and feelings of social isolation [6-9]. To enable safe and functional mobility, a better understanding of how to reliably optimize gait outcomes with RAS in PD is required.

Methods

21 people with idiopathic PD (Hoehn and Yahr stages 2-3, independent walkers) were included in the study and tested during the peak "ON" phase of their medication cycle. Participants were randomized to one of two instruction conditions: free walking (instructed to walk comfortably) or synchronized walking (instructed to synchronize with music). In both conditions, participants walked 8 passes across a 16-foot pressure sensor walkway (Zeno™) to determine baseline gait parameters. Participants then walked to high- and low-groove music that was adjusted to be 10% faster (in beats per minute) than individual baseline cadence. In each condition, participants performed four trials to low groove music (does not induce desire to move), and four trials to high grove music (induces desire to move). Step length, stride width, stride velocity, cadence, and double-limb support time were examined. Coefficient of variation for step length and time were assessed as measures of gait variability.

Results

Overall, high groove cues produced more favourable gait outcomes than low groove cues and, in many cases, metronome cues. Greater step length [F (1.8, 34.7) = 5.19, p = .013, n_p^2 = .22], cadence [F (1.6, 30) = 11.5, p < .001, n_p^2= .38], stride velocity [F (1.7, 32.2) = 11.30, p < .001, n_p^2= .37], and lower double-limb-support time [F (1.5, 27.7) = 7.74, p < .01, n_p^2 = .29] was observed in the high groove compared to low groove condition. In other words, faster gait speed with larger steps and more steps per minute resulted from music that induced desire to move (high groove) compared to music that did not induce desire to move (low groove). In addition, higher stride velocity was observed among synchronized walkers compared to free walkers [F (1, 19) = 7.47, p = .013, n_p^2 = .28], regardless of perceived groove. No significant effects were observed for stride width, step length variability, nor step time variability.

Discussion

These findings support that perceived groove in music and intention to synchronize significantly impact gait outcomes in people with PD while walking to music-based RAS. Larger steps and faster gait are observed with higher groove music, while instructions to synchronize facilitate faster overall gait speed. These findings suggest that controlling for groove and task instructions may foster better, more controlled, clinical outcomes.

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Disclosure Statement

No conflicts to report from any author.

Balance control strategy relationships of upper and lower extremity in two common different style of squat exercise

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Introduction

The main goal of current study was to investigate the relationship between center of mass (COM) and center of pressure (COP) during two different upper body style of squat; hand front of chest (HFC) and hand back of head (HBH) to finding balance and stability characteristics and other biomechanical effect to create balance and stability strategy [1]. Understanding of the movement of the body's center of mass (COM) with respect to the center of pressure (COP) during different style of squat may offer insights into the motor and balance control strategies and provide a basis for approaches that minimize the risk of injuries and recognition biomechanical effect of COP-COM relationship patterns [2].

Clinical Significance

An important question that arises is which (and why) anatomical direction has maximum COM-COP changes when the hands was placed in different upper body positions. Understanding COP-COM relationships reveal clinical aspect of squat exercise executed at two different styles.

Methods

Eleven healthy subjects performed squatting at two different style. The kinematics of the full body joints were reconstructed using a motion analysis system (8 Vicon high speed camera) at 120 Hz. The center of pressure (COP) and center of mass (COM) was obtained using data collected from one force plate at 1200 Hz. MATLAB software 2019b with the BTK Toolkit library and the Mokka (visual biomechanical software) were used to analyze the data. The Shapiro-Wilk test was used to evalate the normality of data, and the t-test was used to compare groups of athletes. When there was no data normality, the Mann-Whitney test was used to compare groups of athletics.The level of statistical significance was set to $p<0.05$ for all tests.

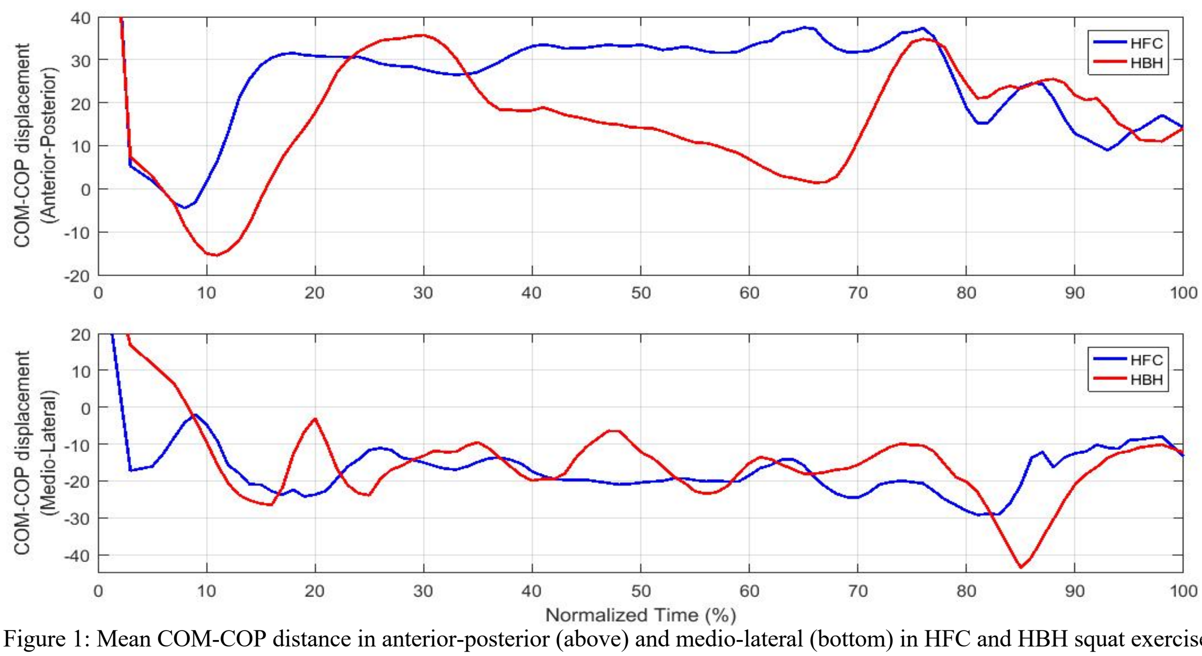

Figure 1: Mean COM-COP distance in anterior-posterior (above) and medio-lateral (bottom) in HFC and HBH squat exercise.

Demonstration

In the upright position there were small changes in the COP and in COM. During the acceleration phase of the squat the COP and COM moved posteriorly, the knee joint torque remained in flexion. During this phase of squatting, the COP and COM returned to the direction of the toe tip, with a significant change of the knee joint moment into extension and the ankle joint moment moving forward the plantar flexion direction.Significant differences between HFC and HBH groups observed at 35-75 percent of squat normalized trial time in anterior-posterior direction ($P<0.05$). As the body went into the deceleration phase, the knee joint moment increased forward extension with major COM and COP distance being observed in both of groups.

Summary

Understanding these kinematic, kinetic and balance strategies during different style of squat (HFC and HBH) is expected to be beneficial to practitioners for using squatting as a task for improving motor function in HFC and HBH. Previous literatures show relationship between COP and COM with joint torque and muscle activation because human motor control change kinetics parameters of joints and muscles with changing COP and COM resulted from change in posture [3]–[5]. According the results, body position generally change the COP and COM relationship and also change the balance control of human movement during squat exercise in HFC and HBH style. In addition, these balance control changes can change muscle roles and activation levels to create motion. On the other hand, the additional flexion of both ankle and knee joints displaced the thigh and shank anteriorly, favoring the displacement of COP towards the direction of the toe tip during the deceleration phase of the squat in HFC and HBH.

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Effects of Eccentric Exercises on Foot Structure, Balance, and Dynamic Plantar Loading

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Introduction

Aging and sedentary lifestyle lead to atrophy of foot muscles.[1, 2] Supervised progressive resistance training have shown to increase toe strength and single leg balance.[3] It is not known if eccentric toe exercise can also alter foot structure and dynamic plantar loading.

Clinical Significance

Healthy foot function is vital for health and well-being. Toe weakness are an independent predictor of falls in older adults.[4]

Methods

Study participants consist of healthy asymptomatic participants, between ages 18 and 40 and without a history of lower limb surgery. Twenty-five subjects provided informed consent and volunteered for the study. Using Repeated Measures study design, participants were evaluated at baseline and following 6 weeks of eccentric home foot and ankle exercises (requiring 10-15 minutes per day, 5 times per week). Exercises consist of warm up, heel raises with knee extended and flexed, calf stretch, posterior tibialis and peroneal muscle exercises – all using ToePro platform with 10-degree tilt that stretched toe muscles. Arch height index, toe grip strength, single leg balance (anterior reach in Y Balance Test) and barefoot plantar pressure during self-selected comfortable pace of walking were assessed at baseline and follow up. Five trials of plantar loading were measured on each foot, using novel emed-X at 100 Hz. The SPSS, version 25 (IBM, Chicago) was used to perform descriptive statistics and normality testing. The lower limb was used as the unit of observation. A Generalized Linear Model with an identity link function was used to test the difference between 2 visits while accounting for potential dependence in bilateral data. The Wald Chi-square was calculated for each dependent variable with significance level of 0.05.

Results

This preliminary results represent data on 16 participants (mean age of 26.8 years and BMI of 23.5 kg/m^2). Participants exhibited moderately pronatory foot posture as indicated by the mean foot posture index composite score of 6.8. Following 6 week of exercises, participants demonstrated a significantly greater toe grip strength (18.3% for hallux and 24.2% for lesser toes), single leg anterior reach (5.1%), and maximum force (2.4%) in barefoot walking. There was no significant change in arch height index, arch height flexibility, and overall foot pronation as measured by center of pressure excursion index. No significant difference was observed in self-selected walking speed as noted by stance time. Exercises also yielded greater plantar pressure in gait but only under hallux (49.9 to 53.7 N/cm^2, p=0.039).

Parameters	Pre	Post	p-value
Arch height index, standing	0.326	0.326	0.864
Arch drop (cm)	0.38	0.35	0.520
Arch height flexibility	13.9	12.4	0.398
Toe grip strength, hallux (kg)	4.21	4.98	0.002
Toe grip strength, toes 2-5 (kg)	3.18	3.95	0.000
Y Balance Test, Anterior (cm)	56.51	59.38	0.007
Stance time, total (ms)	701.7	705.9	0.490
Peak Pressure, total (N/cm^2)	63.5	66.7	0.076
Maximum Force, total (N)	805.1	824.3	0.006
Contact Area, total (cm^2)	124.8	124.3	0.589
Center of Pressure Excursion Index (%)	19.32	19.16	0.793

Discussion

Results suggest that this set of eccentric non-supervised foot and ankle exercises can significantly increase toe grip strength, balance in single leg stance, and maximum force in gait. Exercises also yielded a significantly greater plantar hallucial pressure; this differs from findings of Melai et. al that showed that leg muscle strengthening did not redistribute plantar load in diabetic neuropathic patients.[5] Additional studies needed to investigate the effects of longer intervention, addition of intrinsic exercises, and on different subject population.

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HumanLocomotion.org provided ToePro Exercise Platforms for the study.

Disclosure statement

No Conflict of Interest.

EFFECTS OF VARYING FOAM THICKNESS ON STANDING BALANCE AND ON COGNITIVE TASK IMPACT

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INTRODUCTION

Foam has been one of the more often used passively unstable surfaces (PUSs) in studies of standing balance[1,2]. Among PUSs, foam can be characterized as a non-rigid PUS. Unlike rigid PUSs (such as wobble boards), foam and other non-rigid PUSs (e.g. mud, loose gravel) deform in response to loads imparted by a standing individual's feet.

In previous efforts at this institution, subjects standing on 5 inch thick foam, while instructed to keep the foam surface as still as possible, experienced adverse effects on balance performance[3]. Such findings demonstrate that standing on foam is a more challenging balance task than standing on firm ground; and that this type of non-rigid PUS balance invokes greater use of upper body strategies than fixed surface balance.

However, this work could not assess whether/how the degree of balance task difficulty, for this type of non-rigid PUS, is related to foam thickness. Additionally, the earlier investigation could not evaluate whether any potential impacts of concurrent cognitive tasks on performance or mechanisms, during these non-rigid PUS balance tasks, are affected by foam thickness. The current study investigated the effects, on balance performance and mechanisms, of varying foam thickness during non-rigid PUS balance tasks, as well as the effects of varying foam thickness on performance and mechanism impacts of imposing concurrent cognitive tasks.

CLINICAL SIGNIFICANCE

Understanding how foam thickness mitigates standing balance task difficulty can assist with selecting a thickness to challenge an individual's balance capacity at a desired level. Foam thickness selection can be additionally aided by knowledge of the whether/how varying foam thickness modifies potential concurrent cognitive task effects on performance and mechanisms during this type of non-rigid PUS balance task.

METHODS

Ten healthy subjects (5 m, 5 f; 18-25 years) participated, after providing informed consent. For all trials, subjects were directed to stand on foam, placed atop a fixed force plate (Bertec) and to keep the foam as still as able for 45 seconds. Trajectories for markers, placed in a Helen Hayes arrangement, were recorded (Qualisys) at 100 Hz, while force plate data were synchronously collected at 1000 Hz. Subject performed 2 trials with each permutation of foam thickness (0, 1, 2, 3, 4 & 5 inches), and concurrent cognitive tasks (no cognitive task, NC; verbal fluency task, VF). The primary performance metric (based on task instruction to keep the foam still) was mean foot angular velocity ($Mean_{\omega\_ft}$). Secondary measures were mean angular velocities of pelvis, trunk, upper arms, and lower arms ($Mean_{\omega\_pel}$, $Mean_{\omega\_tr}$, $Mean_{\omega\_ua}$, and $Mean_{\omega\_la}$). Standard deviations of center of pressure in fore/aft and medial/lateral directions ($COP\_sd_{f/a}$ and $COP\_sd_{m/l}$) were also extracted.

RESULTS

With respect to foam thickness, $Mean_{\omega\_ft}$ was significantly greater with 5 in. of foam than with 0-4 inches, and was significantly greater with 4 inches of foam than with 0-1 inches

(Figure 1). With respect to cognitive task, Mean$_{\omega_ft}$ was significantly greater with VF (Figure 1). With respect to foam thickness, Mean$_{\omega_pel}$, Mean$_{\omega_tr}$, Mean$_{\omega_ua}$, and Mean$_{\omega_la}$ did not differ exhibit any significant differences; while with respect to cognitive task, Mean$_{\omega_pel}$, Mean$_{\omega_tr}$, Mean$_{\omega_ua}$, and Mean$_{\omega_la}$ were significantly greater with VF (Figure 2). With respect to foam thickness, COP_sd$_{f/a}$ was significantly greater with 5 inches of foam than with no foam, but did not differ between other thicknesses; and COP_sd$_{m/l}$ was significantly less with no foam, but did not differ between non-zero thicknesses (Figure 1). With respect to cognitive task, both COPs were significantly greater with VF (Figure 1). There were no significant thickness/ cognitive task interactions for any balance performance, secondary or force plate metric.

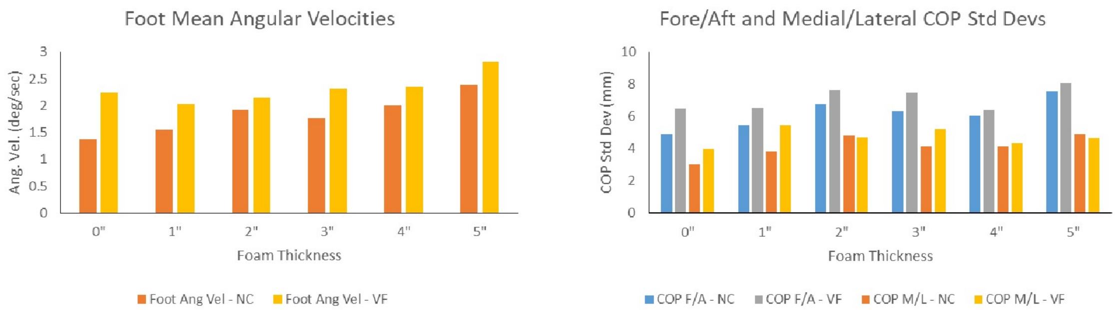

Figure 1. Balance performance measures for foam balance task.

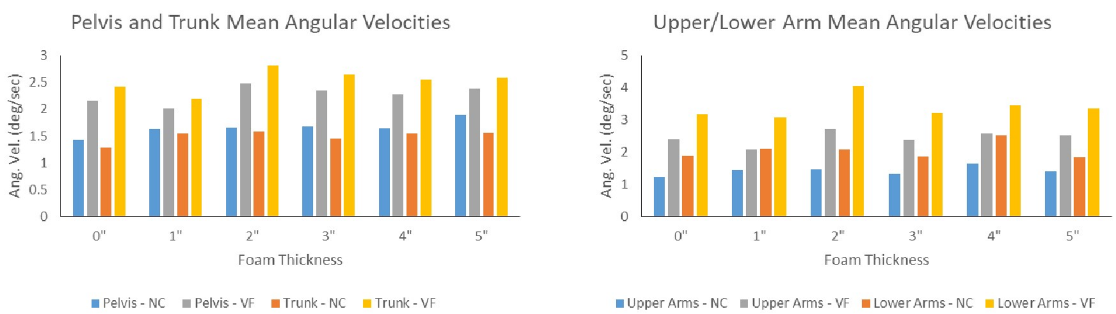

Figure 2. Central body and upper extremity angular velocities for foam balance task.

DISCUSSION

Observed foam effects on foot motion (which corresponds to foam surface movement), in conjunction with lack of any foam thickness effects on central body and arm motions, indicates that foot movement is substantially more responsible for foam induced COP increases. However, substantive influence of the particular foam thickness appears to only be notable between larger thickness increments. Central body and upper extremity motion increases with concurrent cognitive task were consistent with earlier work. Of greater interest, however, lack of significant foam thickness/cognitive task interactions indicate that selection of a particular foam thickness will not interfere with identifying potential impacts of concurrent cognitive tasks

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DISCLOSURE STATEMENT

The authors have no conflicts of interest to disclose.

Relation of Balance, Plantar Tactile Sensitivity, and Fall Incidence in Chemotherapy Induced Neuropathy

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Introduction: Chemotherapy-induced Peripheral Neuropathy (CIPN) has proven to have a negative impact on cancer patients' daily life activities due to its motor and sensory impairments potentially leading to falls and secondary injuries. Studies have shown that the prevalence of peripheral neuropathy among post-chemotherapy patients can range from 30% to 70%, with the highest incidence occurring with patients that undergo treatments with taxanes and platinum derivatives.[1] Peripheral neuropathy motor and sensory symptoms often affect the lower extremities leading to falls and other secondary injuries. However, it is known that loss of balance, muscle weakness and consequent gait disorders are closely related to the high risk of falls for cancer patients with CIPN. Therefore it is the purpose of this study to investigate if there is a relationship between foot plantar pressure, center of pressure excursion, cutaneous sensibility of the foot, and incidence of falls in CIPN patients.

Methods: Three female breast cancer patients (age 68±2 y; body weight 822.4±296 N) after Taxane based chemotherapy reporting CIPN symptoms, and three healthy control (mean age 68 ± 2 y; mean body weight 734.2 ± 177.3 N) matched by gender and age were recruited. Subjects read the research procedures listed in the informed consent approved by Indiana State University IRB and they provided consent. Fall risk was assessed through the Quick Screen Falls Risk Assessment[1]. Sensibility was assessed in five different areas in both feet with the Sorri® Seemens-Weinstein Monofilaments foot kit. Monofilaments with increasing six graduations of stiffness (.07, .4, 2.0, 8.0, 180, 300 g) were sequentially applied until the subject indicated feeling the filament probe and the force associated with that filament was recorded. A medical history was obtained on the medications taken in last 3 months, previous falls in the past year, and vision was tested using a standard vision chart. Peak pressures (P_{Peak}) within the five different areas in both feet and the center of pressure (CoP) excursion were assessed by the Tekscan High Resolution (HR) mat under three different conditions (static balance eyes open; eyes closed and dynamic balance).

Results: Due the limited subject number, descriptive statistical results (mean, median and SD) were calculated for the fall risk, foot plantar sensitivities of the five foot dermatomes, and the foot pressures for the five regions of the foot during static balance standing (eyes open & closed) and dynamic balance (walking). The mean fall risk assessed by the Quick Screen Falls Risk questionnaire for the chemotherapy-induced peripheral neuropathy (CIPN) subjects was 7.88±4.15 and the control group was 1.50±0.56. The foot dermatome pressure sensitivities for the right and left feet for the CIPN and control subject groups are graphically presented in Figure 1.

The results were consistent with previous studies with diabetic peripheral neuropathy having correlated the loss of sensibility in the fore-foot and rear-foot areas to an increased pressure in those areas, muscle imbalances and decreased proprioception. The CIPN group's anterior-posterior (A-P) CoP excursions for standing with eyes open were 4.57±1.57cm for the breast cancer CIPN group, CIPN group eyes closed 6.0 ± 0.6cm and the control group eyes open condition was 3.1 ± 1.2cm and eyes closed 3.4 ± 0.5cm. Dynamic balance during walking with the Tekscan foot pressures measured found that the plantar foot pressures were higher in the forefoot of the cancer survivors, whereas the control group showed in greater weight concentration to the rearfoot and is shown in Figure 2.

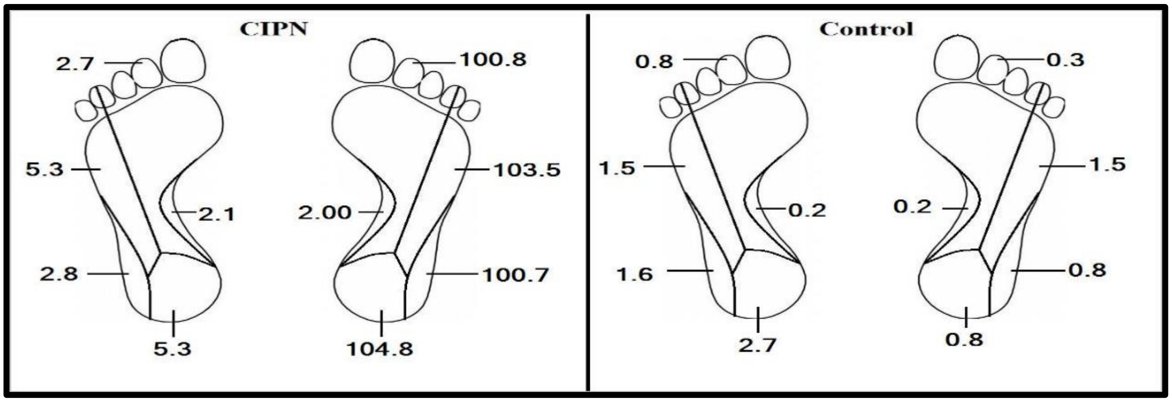

Figure 1. Foot region dermatome sensitivity in grams for CIPN and control subjects

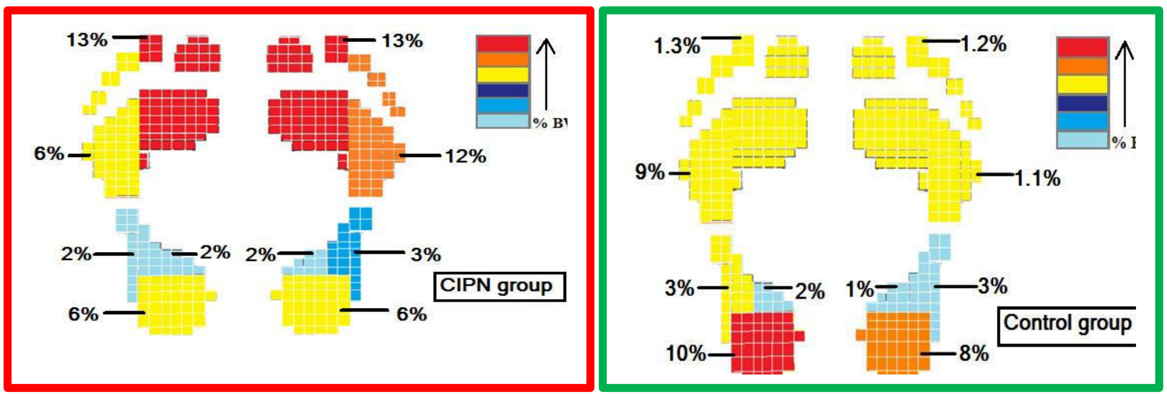

Figure 2. Plantar foot pressures while walking in cancer survivors and control group

Discussion: The standing balance CoP excursion results indicated that cancer patients experienced 47% more A-P motion using visual, vestibular, and proprioceptive feedback, and when their eyes were closed which would utilize vestibular and proprioceptive feedback they cancer survivors experienced 75% more A-P motion while static standing. The walking plantar pressures for the cancer group were elevated in the forefoot as compared to the control groups and this could be a contributing factor to forward toppling and increased risk of falls. This was consistent with the finding of patients who were exposed to neurotoxic chemotherapy agents had poorer static and dynamic postural control.[2]

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Long-term outcomes of intensive versus minimal spasticity management strategies on ambulatory individuals with cerebral palsy: preliminary results of a multicenter study

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INTRODUCTION: Spasticity management strategies in cerebral palsy (CP) differ widely. Selective dorsal rhizotomy (SDR) is the core of an intensive spasticity management approach. However, the quality of past long-term SDR outcomes research in CP is poor.[1,2] The aim of this study is to compare the long-term outcomes (>10 years) of an intensive spasticity management strategy, primarily by means of SDR, to a group receiving minimal spasticity management.

CLINICAL SIGNIFICANCE: Comparing outcomes between two groups who were matched at baseline but underwent different intensities of spasticity management during childhood provides comprehensive insight into the long-term effects of clinical decisions for this population.

METHODS: The design of this study leveraged two distinct spasticity management approaches that differed by institution. Detailed baseline matching and inclusion criteria were previously described.[3] This multicenter cohort included two groups: individuals treated with intensive spasticity management, which included SDR as the primary treatment during childhood (Yes-SDR), and individuals with minimal spasticity management and no history of SDR (No-SDR). Participants had bilaterally involved spastic CP and were retrospectively matched on lower limb spasticity profile (LSP) at baseline.[3] LSP ranges from 0 to 5, where 5 represents the highest spasticity profile. All potential participants were categorized by their LSP and then sampled in a manner that reflected the distribution of baseline spasticity among SDR-eligible patients. Study individuals ≥21 years old were invited to return for a follow-up visit to evaluate gait kinematics, walking energy expenditure, gross motor function, and physical examination measures.

RESULTS: Seventy-one participants were included in this preliminary analysis (Yes-SDR, n=34 [85% of recruitment target]; No-SDR, n=37 [93% of recruitment target]). The Yes-SDR group was younger at the time of baseline gait analysis (Yes-SDR = 5.2±1.1 years, No-SDR = 8.2±1.9 years; $p<0.01$). Age at SDR was 5.7±1.2 years. The time between baseline and follow-up for the Yes-SDR group was longer than the No-SDR group (Yes-SDR = 21.2±2.6, No-SDR = 17.4±3.1 years; $p<0.01$). Groups were, on average, the same age at follow-up (Yes-SDR = 26.4±2.2 years, No-SDR = 25.9±3.3 years; $p=0.24$).

<u>Lower Limb Spasticity Profile (LSP)</u>:[3] Average LSP was retrospectively matched at baseline (Yes-SDR = 2.5±0.6, No-SDR = 2.3±0.6; $p=0.14$). The LSP of No-SDR was significantly higher at follow-up (Yes-SDR = 1.1±0.1, No-SDR = 2.2±0.6; $p<0.01$).

Gait Deviation Index (GDI):[4] At long-term follow-up, there was no significant difference in overall gait quality between groups (Yes-SDR = 69.9±10.8, No-SDR = 70.6±10.6; p=0.77; Figure 1).

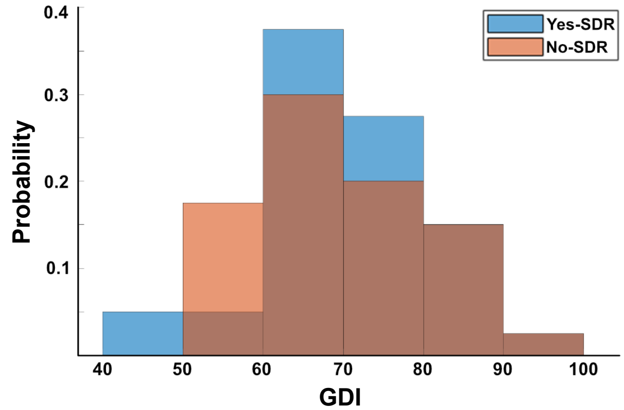

Figure 1. *Probability of the Gait Deviation Index (GDI) in both groups at long-term follow-up.*

Energy Expenditure: At long-term follow-up, there was no significant difference in net non-dimensional walking energy consumption (Yes-SDR = 0.144, No-SDR = 0.131; p=0.20).

Motor Function: (a) Gillette Functional Assessment Questionnaire (FAQ):[5] Median FAQ scores were similar for both groups (Yes-SDR = 9.0 [4-10], No-SDR = 8.5 [4-10]; p=0.20). (b) Gross Motor Function Measure (GMFM):[6] There were no differences observed in dimension D: standing (GMFM-D; Yes-SDR = 76.3±18.1; No-SDR = 75.7±15.4; p=0.89), dimension E: walking, running and jumping (GMFM-E; Yes-SDR = 63.5±29.3; No-SDR = 61.5±23.0; p=0.75), or the 66-item total score (GMFM-66; Yes-SDR = 69.1±12.0; No-SDR = 68.1±7.8; p=0.67). Results of GMFM-D, GMFM-E, and GMFM-66 were further stratified into three groups according to Gross Motor Function Classification System (GMFCS) level (I vs. II vs. III+IV) and no statistical differences were observed between the Yes-SDR and No-SDR groups (p>0.10).

DISCUSSION: Our preliminary data show that differing spasticity management strategies (intensive vs. minimal) had similar effects on gait kinematics, walking energy expenditure, and gross motor function at long-term follow-up. These similarities between groups were observed, despite the Yes-SDR group having significantly lower spasticity than the No-SDR group at follow-up. The present study suggests that early intervention and elimination of lower limb spasticity via SDR does not translate to significant functional advantages in adulthood.

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Genu Valgum Following Distal Femoral Extension Osteotomy in Children with Cerebral Palsy

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INTRODUCTION

Crouch gait is a common problem among children and adolescents with spastic diplegic cerebral palsy (CP) [1]. Due to the combination of knee flexion contracture and quadriceps insufficiency, surgical intervention is often recommended to maintain long term ambulatory function and decrease pain [2]. A distal femoral extension osteotomy (DFEO) can be indicated to restore full extension to the knee joint in patients with crouch gait [3]. However, the effect of DFEO on physeal growth arrest in skeletally immature patients remains unclear. Physeal tethering may occur due to the proximity of the plate to the distal femoral physis, which can result in a coronal plane angular deformity. The purpose of this study was to (1) assess the extent of any coronal plane angular deformity that developed secondary to surgery and (2) determine if distance between the osteotomy and distal femoral physis influenced the severity of deformity.

CLINICAL SIGNIFICANCE

Sagittal plane deformity in patients with crouch gait following DFEO has been previously described [4]. However, analysis of angular deformity in the coronal plane in skeletally immature patients with open physes has not been conducted.

METHODS

A total of 78 limbs from 48 patients diagnosed with spastic cerebral palsy who underwent DFEO at our institution between 1999-2016 were included in this retrospective study. Anatomic lateral distal femoral angles (aLDFA) and distance between hardware and distal femoral physis were measured from radiographs pre-operatively and at final follow-up up to 36 months following surgery and analyzed using linear mixed model regressions. Demographics and other relevant clinical information such as Gross Motor Function Classification System (GMFCS) level were also recorded.

RESULTS

The average age at time of surgery was 13.0 ± 2.2 years. 12 limbs (15.4%) were from patients of GMFCS level I, 25 (32.1%) were level II, and 41 (52.6%) were level III. aLDFA measurements progressed towards valgus alignment during the 36 months following surgery (-0.77° per year, $p=0.0021$) (see Fig. 1). This effect was more significant for younger patients (estimated -5.08° per year for an 8-year-old patient, p=0.0016; -0.66° per year for a 14-year-old patient, p=0.2069). However, no patients required a revision procedure to correct angular deformity. There was a moderate association between distance from hardware to physis and the change in aLDFA. Although not statistically significant, plates closer to the physis were associated with an increase in valgus alignment (-0.06° change in aLDFA for every 1 mm decrease in distance, p=0.4169) (see Fig 2.). The median time from surgery to hardware removal was 1.4 years (interquartile range: 0.5 to 4.8 years).

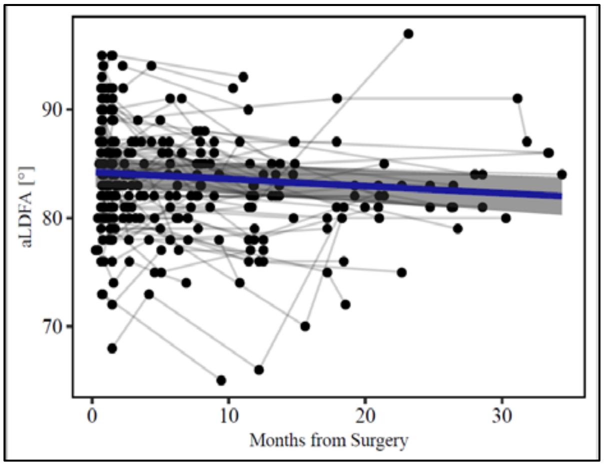

Figure 1. Progression of aLDFA following DFEO.

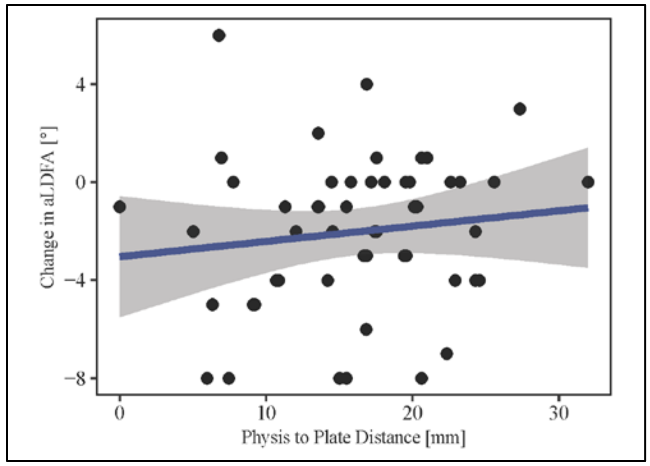

Figure 2. Change in aLDFA as a function of distance between distal femoral physis and blade plate.

DISCUSSION

These findings suggest the DFEO remains a safe intervention for treatment of crouch gait that does not induce a clinically significant distal femoral coronal plane abnormality. We found distance between blade plate and physis is not a significant predictor of valgus deformation. Age at time of surgery is an important factor to consider when assessing risk of valgus progression following DFEO. Surgery in skeletally immature patients is safe coupled with prompt hardware removal but younger patients should be closely monitored for secondary complications.

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DISCLOSURE STATEMENT

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Instrumented Gait Analysis for the Clinical Management of Children with Cerebral Palsy: A Scoping Review

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INTRODUCTION

Although use of Instrumented Gait Analysis (IGA) for the clinical management of individuals with cerebral palsy (CP) has increased in recent years, there is significant variability in data collection protocols, as well as reporting and interpretation strategies. Appreciating this variability is difficult with an ever-growing body of literature. Previous investigators have performed systematic reviews of the literature to evaluate and summarize the evidence related to the efficacy of IGA. Wren et al (2011) identified 1528 manuscripts related to IGA and categorized them based on a framework used to evaluate the efficacy of diagnostic tests (Fryback and Thornbury, 1991). This comprehensive work included applications of IGA for all patient populations. However, children with CP present with unique clinical presentations and gait characteristics that affect data collection, and the methods of reporting and strategies for interpretation justify a more focused review of the literature. Therefore, the purpose of the current work was to perform a scoping review of the existing literature to describe and categorize the range of existing literature about IGA as applied to the clinical management of children with CP.

SIGNIFICANCE

This scoping review will provide clinicians and clinical researchers with a resource to summarize the roles of IGA in the clinical management of children with CP. Results will be beneficial to clinical researchers by identifying gaps in the existing scientific literature. This review was undertaken in preparation for development of a clinical practice guideline (CPG) on this topic.

METHODS

A health sciences librarian developed the search strategy with 4 key inclusion criteria: a) original peer-reviewed research study; b) population included children with CP; c) used IGA to investigate gait; and d) available in English. Records were identified by a search of Medline (via Ovid) completed in April 2019. Once studies were identified, to ensure reliability, individual sets of 10 titles and abstracts were screened by all 6 reviewing authors for inclusion. After 100% agreement was achieved between the 6 reviewers, individual investigators screened the remaining studies. The included studies were classified into nine categories developed based on Wren et al.'s 2011 systematic review about IGA (Table 1.0). To ensure reliability of categorization, individual sets of 10 full text articles were categorized by 6 authors. After 100% agreement was achieved, individual investigators categorized the remaining studies. To describe the methodological characteristics of the existing literature about IGA and how it has been used for the clinical management of children with CP, frequency (percentages) of the categories was then calculated.

Table 1. Study Categories
1. Reliability and validity testing of IGA or IGA measures relative to existing accepted measures
2. Validation of IGA measures for diagnosis and/or Identification of sub-groups via IGA analysis (Incudes documenting kinematic patterns using IGA)
3. Effectiveness of IGA or IGA measures to determine treatment approach
4. Effectiveness of using IGA or IGA measures to improve patient outcomes
5. Cost-effectiveness or cost-benefits of using IGA or IGA measures within clinical care
6. IGA or IGA measures used as a tool to investigate the effectiveness of some intervention
7. Commentary or review
8. Did not involve gait analysis or did not use IGA or did not include children with CP
9. Insufficient information available in English

RESULTS

1296 citations were screened, and 599 included studies were categorized. Analysis of study purposes and designs showed a wide range of prospective and retrospective designs analyzing a myriad clinically important gait features (Table 2.0). The most common study designs included those that develop/evaluate the clinimetric properties of quantitative metrics used to evaluate the gait of children with CP and those that identify specific atypical gait characteristics of children with CP. The least commonly reported categories included studies that evaluated the cost-effectiveness or cost-benefits of IGA.

Table 2. Category frequency of included studies (Total number of studies less 43 studies in dual categories: N=599)

Category	Frequency N (%)
1,2	315 (53%)
6	298 (50%)
3, 4, or 5	29 (5%)
Dual Categories	43 (7%)

DISCUSSION

As a key technology for documenting gait dysfunction, IGA provides the basis for a wide range of approaches to the examination, evaluation, and measurement of outcomes for gait dysfunction, as well as forming a crucial step in clinical decision making for treatment for children with CP related gait dysfunction. By identifying and characterizing a substantial body of peer-reviewed research on these topics, this scoping review has begun to describe the spectrum of existing literature on IGA for children with gait dysfunction related to CP. As such, this scoping review can form the basis to begin developing Clinical Practice Guidelines (CPG) about the use of IGA for the clinical management of children with CP. Additional steps such as a search of additional databases and analysis of the studies identified, stakeholder surveys, development of precise PICO(T) questions for the CPG and quality assessment of the literature are progressing under sponsorship of the Academy of Pediatric Physical Therapy.

Wren TA, Gorton GE, 3rd, Ounpuu S, Tucker CA. Efficacy of clinical gait analysis: A systematic review. Gait & posture. Jun 2011;34(2):149-153.

Dennis G. Fryback, and John R. Thornbury. (1991). The efficacy of diagnostic imaging. Medical Decision Making, 11, 88-94.

DISCLOSURE STATEMENT - We have no conflicts of interest to disclose.

Factors Associated with Walking Activity in Adults with Cerebral Palsy

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Introduction: This IRB approved prospective study uses instrumented gait analysis, patient reported outcomes (PROs), and portable accelerometers to examine walking activity (WA) and factors with associated WA.in adults with cerebral palsy (CP).

Clinical Significance: Previous studies have looked at WA in children with CP, but few have explored factors that influence WA in adulthood using gait analysis and patient reported outcome measures. The aim of this work is to inform pediatric and adult specialty practice by providing objective data of levels of WA and factors that influence activity in adults with CP.

Methods: We identified adults with CP who in 2017 were between the ages of 25-45 and had a previous gait analysis at our institution. 109 participants returned to the lab to complete a 3D gait analysis, self-report outcomes (PROMIS physical function, Satisfaction with Life Score (SWLS) and demographic information), and wear a StepWatch™ to monitor community WA. The StepWatch™ is a device that fits around the ankle and measures the number of strides taken by that leg in a day. Average stride data based on GMFCS classification was compared with a group of 193 nondisabled adults (NDA) age 30-39 years [1] utilizing Welch's t-tests with Bonferroni corrections. We evaluated correlations, stratified by GMFCS level, between WA and gait deviation index (GDI), gait speed, level of physical function (PROMIS), quality of life (SWLS), Body Mass Index (BMI), employment rate, and caretaker need.

Results: 109 adults with CP, age 29±4 years, returned to participate. GMFCS levels were: 20% level I, 54% level II, 22% level III, and 4% level IV. PROMIS physical function, GDI, and gait speed had positive correlations with strides/day and showed some significant differences compared to adults without disability (Tables 1 & 3). WA (strides/day) was moderately correlated with GMFCS level (r = -0.63). Compared to nondisabled adults, WA was higher in the GMFCS I group, no different in the GMFCS II group, and lower in the GMFCS III and IV groups (Table 1). Employment and caretaker needs had small correlations with GMFCS level (r = -0.32 and r= 0.35), while quality of life score and BMI had less association with GMFCS level (Table 2).

GMFCS	NDAs	I	II	III	IV
WA (strides/day)	2564(1417)	4324(2003)*	2705(1568)	890(569)*	130(33)*
PROMIS physical Function	55(8)	50(9)	41(9)*	34(10)*	32(5)
GDI	100(10)	85(10)*	73(11)*	64(10)*	55(15)*
Gait Speed	124(18)	113(12)*	82(21)*	48(22)*	27(14)*

Table 1: Average strides/day, Gait pattern/velocity and self-reported physical function by GMFCS level compared with nondisabled adults. The mean and standard deviation (SD) are

shown. A p-value of <0.0125 denotes a statistically significant value due to 4 comparisons with Bonferroni correction. Statistical significance is marked by an asterisk (*).

GMFCS	NDAs	I	II	III	IV
% Employed	96	83	54	38	25
% with Caretaker	n/a	8	31	46	80
Average SWLS	20-24 (Average)	27(7)	24(7)	23(8)	23(8)
Average BMI	22(2)	25(8)	26(7)	28(6)	29(4)

Table 2: percent employment, percentage with caretaker needs, SWLS, and BMI. As of Oct. 2019, the US unemployment was 4% [3]. The SWLS score range for nondisabled adults is shown [2]. The BMI of a NDA adult is from a normal weighted adult living a mostly sedentary lifestyle [4].

Variable	r with WA (strides/day)	p value
PROMIS (Physical Function)	0.42	6.4e-6 *
GDI	0.48	1.2e-7 *
Gait Velocity	0.58	2.8e-11 *
Employment	0.27	0.0046 *
Care	-0.22	0.020
SWLS	0.22	0.029
BMI	-0.17	0.078

Table 3: Correlations between WA (strides/day) and gait and self-reported factors with Bonferroni corrections (significance* at $p < 0.0071$).

Discussion: WA in adults functioning at GMFCS I or II met or exceeded the WA of an age-matched nondisabled sample. Clear association was noted between physical function and walking activity as there was a significant correlation between strides per day and the PROMIS physical function score. Adults functioning at GMFCS levels III and IV took significantly fewer strides per day compared to nondisabled adults. Significant associations were found between walking activity and employment. Despite limited physical function and the presence of gait deviations, adults with CP demonstrated levels of satisfaction with life that were no different from nondisabled adults.

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REPEATABILITY OF MULTI-SEGMENT FOOT KINEMATICS IN PEDIATRIC PATIENTS WITH CEREBRAL PALSY

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INTRODUCTION

The reliability of three-dimensional gait kinematics has been reported for lower body segment and joint angles using traditional single segment foot models.[1, 2] However, with the increasing use of multi-segment foot models, it is important to understand the repeatability of these models. As such, the repeatability of the IOR multi-segment foot model has been assessed in healthy adults[3] and in adults with foot pathology.[4] However, there is little information about the repeatability of this model in pediatric patients with cerebral palsy. Therefore, the purpose of this study was to determine the repeatability of the IOR foot model in pediatric participants with cerebral palsy.

CLINICAL SIGNIFICANCE

The variability in multi-segment foot kinematics, both within and between therapists, is an important consideration when using multi-segment kinematics as part of a clinical decision-making process.

METHODS

The study was approved by the Baylor College of Medicine institutional review board. Five pediatric participants with cerebral palsy were recruited and participated in the study. Each participant completed six data collections total, three data collections by each of two physical therapists. Both physical therapists are board-certified clinical specialists in pediatric physical therapy. They have 6 and 9 years of clinical experience in 3D motion analysis and 2 and 1.5 years' experience, respectively, with the IOR multi-segment foot model. During these data collections, shank and foot kinematics were collected using the IOR multi-segment foot marker set and model.[5] Kinematic data were collected at 120 Hz with a 12-camera motion capture system (Vicon, Oxford, UK). Marker trajectories were filtered with a low-pass Butterworth filter with a cut-off frequency of 6 Hz. All kinematic data processing was performed in Visual 3D (C-Motion, Inc., Germantown, MD). Data from six strides per data collection session for each participant were used for analysis. In order to quantify repeatability, the standard error of measurement (SEM) for each joint angle was calculated across both therapists (overall SEM).

RESULTS

The data from five participants (2 male and 3 female) ages 11.7 ± 5 years old were analyzed. Overall SEM values across both therapists were < 5 degrees for 10 of the 15 joint angles and ≤ 7 degrees for 100% of the joint angles (Fig. 1). Overall SEM was lowest in the coronal plane and highest in the sagittal plane.

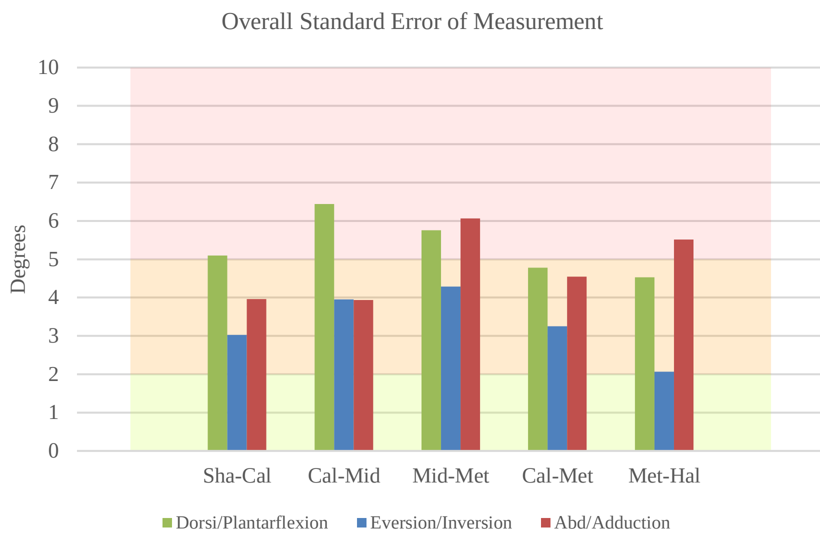

Figure 1. Standard Error of Measurement (SEM) in degrees across both physical therapists for the IOR multi-segment three-dimensional relative segment angles.

DISCUSSION

The results of the current study are consistent with the repeatability of the IOR foot model reported for adults. For example, in a repeatability study of the IOR foot model on healthy adults, Deschamps et al. reported absolute angle values between 2.8 to 7.6 degrees, which are consistent with the results of our study.[3] When assessing repeatability in adults with foot deformity using the IOR foot model, Deschamps et al. found an increase in absolute error values and particularly higher variability in the 3D rotations that involved the midfoot.[4] Even though the bony landmarks of the foot can be more challenging to palpate in pediatric patients with cerebral palsy, the results of this current study demonstrate that it is possible to achieve similar levels of consistency as that obtained in healthy adult patients. These results provide evidence that the IOR foot model can be adequately applied for clinical use in a pediatric patient population as long as the SEM is considered during data interpretation.

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DISCLOSURE STATEMENT

Authors have no conflicts of interest to disclose.

Center of Mass Excursions are Exaggerated and Delayed in Children with Cerebral Palsy During Visual Fall Perturbations while Walking in a Virtual Reality Environment.

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INTRODUCTION

Deficits in the control of balance and movement are primary impairments for children with cerebral palsy (CP). While traditional rehabilitation interventions in CP are motor-centric and have focused on improving function by reducing motor deficits associated with muscle spasticity and strength, recent studies[1,2] have suggested that a significant contributor to the motor deficit in CP may be impairment in the neural processing of sensory information, as evidenced by disrupted thalamocortical tracts and impaired somatosensory cortical activation in children with CP. These findings highlight a need for assessing the balance response to sensory perturbations while walking in children with CP. We proposed to investigate this by assessing the balance response to a visual fall stimulus in the form of a sudden tilting of a virtual reality environment in the frontal plane, while children with and without CP walked on a treadmill. Our hypothesis, based on our previous study on typical adults[3], was that children with CP would demonstrate delays or differences in coordination between multiple balance mechanisms, or an absence in specific recovery mechanism compared to typically developing (TD) children.

CLINICAL SIGNIFICANCE

Characterizing the differences in balance strategies used by children with and without CP in response to sensory perturbations is important to enable development of targeted, sensory-centric therapies that address the CP-specific walking balance deficits.

METHODS

Four subjects with CP between the age of 8-18 years (2 with hemiplegia, 2 with spastic diplegia), Gross Motor Function Classification System (GMFCS) level I-II, were recruited along with four age-matched controls. Subjects walked on a self-paced treadmill in a virtual reality environment. They received a visual fall stimulus, triggered on heel strike of either foot, with 12-13 steps intervening steps between stimuli. The direction of the fall stimulus was towards the side of the triggering heel-strike, and randomly determined triggers without stimulus served as controls in the analysis. Each walking trial lasted two minutes, and was repeated ten times, with intermittent rest breaks. Prior to the data collection trials, individuals participated in two bouts of two-minute practice trials to acclimatize to the self-paced treadmill and virtual reality environment. Our primary outcome variables were center of mass (CoM) excursion and velocity.

RESULTS

Subjects in both groups reacted to the visual fall stimuli by moving their CoM away from the direction of the perceived fall. Children with typical development (TD) reached the peak average CoM excursion of 5.2±1.7 cm in away direction typically at the 4^{th} post-stimulus step while children with CP demonstrated a more magnified but delayed shift in CoM in the away direction, with the peak average CoM excursion of 8.3±3.3 cm in the away direction occurring

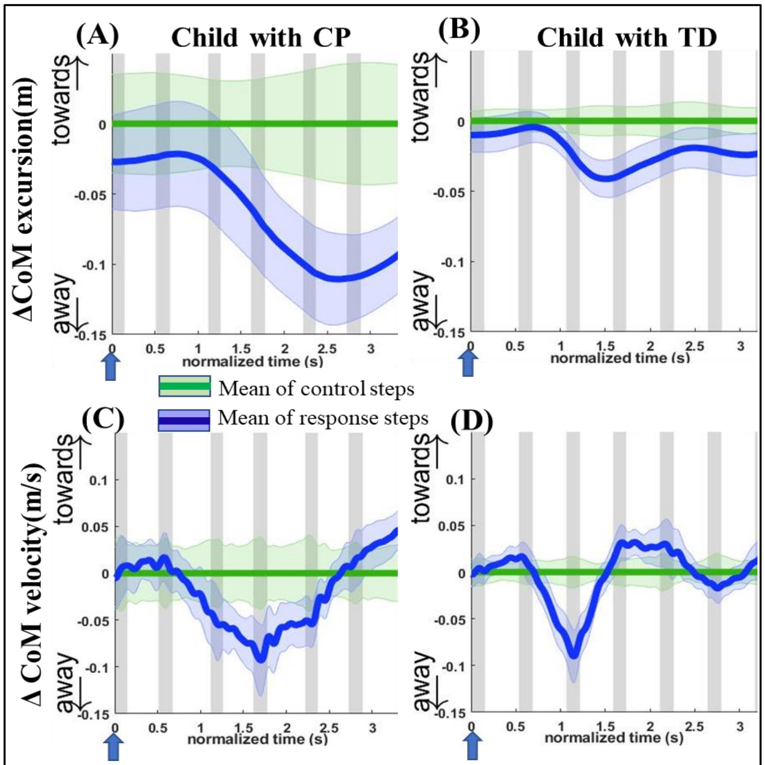

Fig. 1. Medio-lateral center of Mass (CoM) excursion and velocity trajectories in response to visual fall stimuli in one representative subject with CP **(A & C)** and with TD **(B & D)** respectively. Green lines indicate the mean of control (no fall stimulus) steps, which is subtracted from stimulus data, and resulting response is depicted as blue lines. Shaded areas around each line represent one standard deviation. Y axis shows 6 steps, time-normalized to 100 timepoints, with double-stance periods shaded gray and single-stance periods white. Arrows at the beginning of each curve mark the triggering visual fall stimulus.

	CoM peak excursion(m)		CoM peak velocity(m/s)	
	Mean	SD	Mean	SD
TD	0.05	0.02	0.03	0.02
CP	0.08	0.03	0.04	0.02
Effect Size	1.36		1.03	

Table 1. Mean, standard deviation (SD) and effect sizes for CoM excursion (meters) and velocity(meters/sec).

typically on the 5th post-stimulus step (**Fig 1**). The average peak CoM velocity for the TD group peaked during the 3rd post-stimulus step while that for the CP group peaked during the 4th post-stimulus step. The magnitudes of both peak average CoM excursion and velocity were higher in CP vs. TD group, as evidenced by large effect sizes (**Table 1**).

DISCUSSION

Our results suggest that compared to typically developing peers, our cohort of children with CP demonstrated a delayed but magnified balance response to sensory perturbations while walking. This might be suggestive of delayed neural processing and poor sensory integration abilities in children with CP. Previous studies[4] suggest there is a shift from visual dominance towards kinesthetic inputs while mediating postural responses by 6 years of age. A magnified CoM response in an older cohort of children with CP may be indicative of over-reliance on vision, a characteristic seen in younger TD, thus suggesting that CP group's response may not have matured to the level of TD peers. Future research efforts with larger sample size and a wider age range for both CP and TD groups is needed to provide stronger support to the findings of this pilot study.

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DISCLOSURE STATEMENT

The authors have no conflicts of interest to disclose.

Long-term outcomes of femoral derotation osteotomy in individuals with cerebral palsy

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INTRODUCTION

Individuals with cerebral palsy (CP) often have abnormal femoral anteversion, internal hip rotation or intoeing during gait. An external femoral derotation osteotomy (FDO) may normalize hip rotation in the short- and long-term, though previous outcome studies have design limitations to objectively assess its potential benefit compared to no FDO [1-4]. The purpose of this study is to address these limitations by assessing long-term outcomes in individuals with CP who had an FDO in childhood compared to matched-individuals who did not have an FDO. The secondary purpose was to compare outcomes within groups. Outcome assessments were guided by the domains of the International Classification of Functioning, Disability, and Health. This abstract focuses on body structure and function outcomes.

CLINICAL SIGNIFICANCE

There is a paucity of evidence as to whether childhood surgeries improve or maintain function, activity, and participation into adulthood compared to a control group. These long-term outcomes can help clinicians more comprehensively counsel families.

METHODS

For this cohort study, eligible individuals were identified within our gait lab database who met the following criteria: 1) diagnosed with bilateral CP, 2) had an FDO when 5-12 years old (10-90th percentile for historic patients), 3) had a pre-operative gait analysis (Pre) or a gait analysis at which they met all other criteria (non-FDO group), and 4) currently ≥25 years old. The non-FDO group was matched to the FDO group based on their anteversion and Pre hip rotation in gait.

Participants returned for a Long-term analysis, which included various measures from gait analysis, physical exam, radiographs and functional tests. Statistics were performed using Matlab. The non-FDO group only included individuals in GMFCS levels I-II (determined at Long-term), so between-group comparisons were only performed for these individuals.

RESULTS

Sixty-one individuals participated (50 FDO, 11 non-FDO; Table). Groups were matched at Pre ($p \geq 0.083$) on all variables except hip abductor moment ($p=0.048$; Figure). Anteversion by physical exam, mean stance hip rotation, and Tonnis angle improved in all groups ($p \leq 0.047$). Reimer's Index only improved in the FDO group ($p=0.003$). Hip abductor moment did not change in any group ($p \geq 0.310$). At Long-term, there were no differences between the FDO and non-FDO group in hip rotation, hip abductor moment, number of hip abductor repetitions, Reimer's Index or Tonnis angle ($p \geq 0.657$). Anteversion was smaller in the FDO group ($p=0.016$).

Table. Demographics of participants (mean (SD) [min-max])

				GMFCS level			
Group	**Pre Age (yrs)**	**Long-term Age (yrs)**	**Sex (M/F)**	**I**	**II**	**III**	**IV**
FDO	8(2) [5-12]	29(3) [25-35]	25/25	16	17	13	4
Non-FDO	9(2) [5-11]	28(4) [25-36]	4/7	4	7	0	0

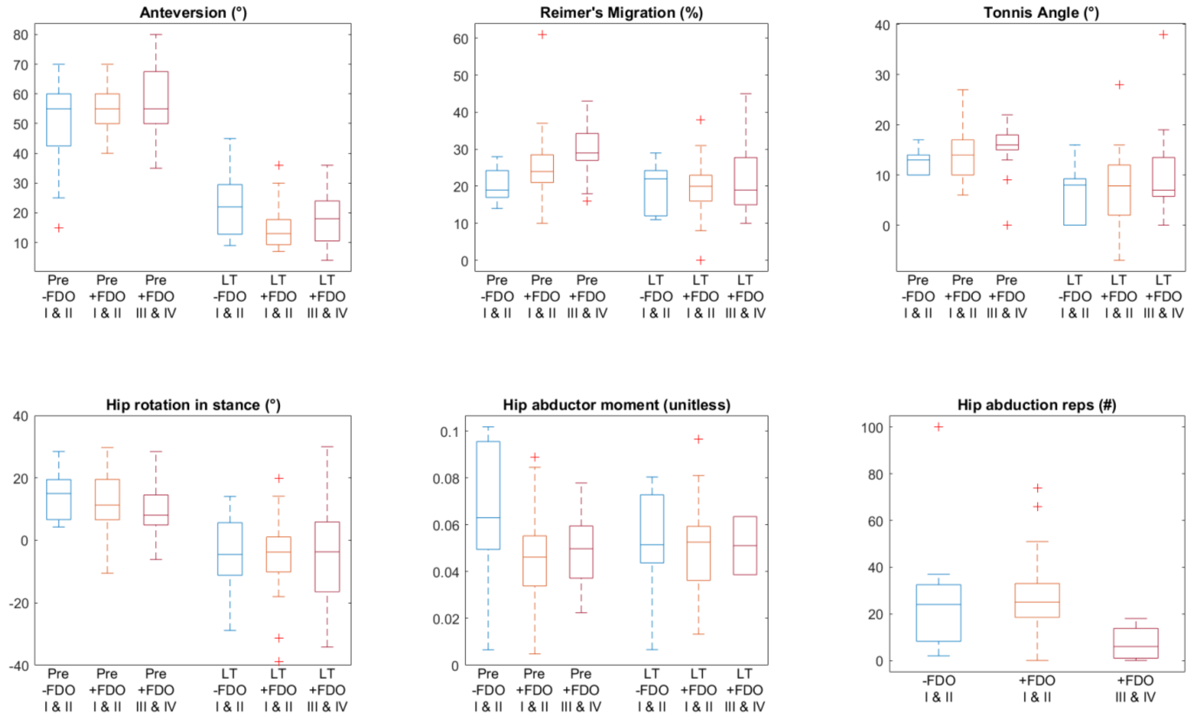

Figure. Outcomes for the different FDO and non-FDO groups at the baseline (Pre) and Long-term (LT) assessment. GMFCS levels (I-II; III-IV) are plotted as different colors.

DISCUSSION

Both groups had decreased hip rotation and anteversion at Long-term compared to Pre, in agreement with our retrospective study [4]. This suggests natural femoral remodeling (non-FDO) and maintenance of surgical correction (FDO group). GMFCS level I-II groups were matched at Pre on the radiographic measures, so the small improvements in Reimer's may be due to the FDO, whereas improvements in Tonnis angle are not. Radiographic improvements are also evident in individuals in GMFCS levels III-IV. Normalized anteversion theoretically improves frontal plane hip abductor moment arm; however, this was not reflected in the data. The groups had similar hip abductor moment and performed a similar number of hip abductor repetitions.
In summary, an external FDO results in lasting normalization of anteversion, which exceeds improvements without an FDO. However, the data suggest that an FDO may not be necessary for patients in GMFCS levels I-II if their goal is to maintain or improve functional parameters, such as hip rotation, abductor moment, or work capacity. Alternatively, an FDO will help patients achieve neutral hip rotation quicker.

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

Abnormal plantar flexor force and work patterns underlying equinus gait

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INTRODUCTION

Children with cerebral palsy (CP) often exhibit contractures, spasticity and abnormal motor control [1], which can collectively contribute to inappropriately timed and modulated muscle forces during walking. Conventional gait analysis lacks the specificity needed to infer the muscle-tendon forces underlying gait disorders. We are investigating the use of a new sensor technology – shear wave tensiometery – to directly assess muscle-tendon forces during walking [2]. In this study, we used tensiometry to evaluate plantar flexor force patterns in children with CP who exhibit equinus gait (toe walking). We also coupled the tendon force data with kinematics to characterize mechanical work loops of muscle-tendon units, providing a quantitative assessment of functional slack lengths, effective stiffness, and work production in gait (Fig. 1).

CLINICAL SIGNIFICANCE

Treatments for gait disorders in children with CP, e.g. spasticity treatments, muscle lengthening procedures, and tendon transfers, are intended to alter muscle-tendon actions in gait. However, conventional gait analysis cannot be used to infer the muscle-tendon actions that induce observable joint kinetics and kinematics. This study demonstrates the use of shear wave tensiometry to quantitatively assess muscle-tendon behavior during gait, which could enable more objective treatment planning and outcome evaluation for gait disorders.

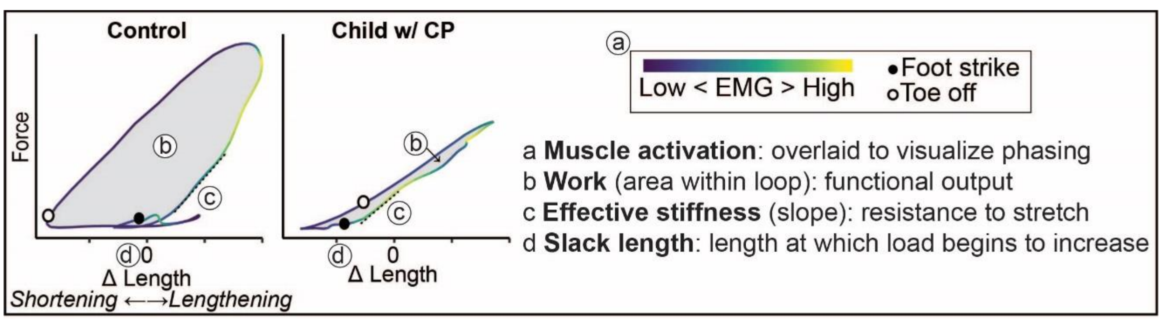

Figure 1. Representative work-loop of the plantar flexors about the ankle. Shown are measurements from a typically developing child (control) and child with (w/) CP who exhibited equinus gait . This comparison illustrates the early onset activation, shorter slack length, reduced effective stiffness and substantially diminished work production that can underly equinus gait.

METHODS

Two children with CP (2M, 14 and 16 years) and seven typically developing children (controls) (5M/2F, 8 to 16 years) have participated to date. One child with CP (14 years) received a posterior selective rhizotomy (L2, S1) 17 months prior to testing. The other received a right gastroc-soleus lengthening (Baker) 6 years prior, and a multi-level surgery – including a right limb internal derotation osteotomy – 3 years prior to testing. Reflective markers and EMG sensors were placed bilaterally over the lower limbs of all participants [3]. A shear wave tensiometer was secured over the Achilles tendon of the more affected limb of the children with CP at the time of testing, as defined by spasticity scores, and on the right limb in the controls. A subject-specific calibration procedure was used to estimate Achilles tendon force from shear wave speed [4]. Subjects walked overground at their preferred speed. A minimum of five strides were analyzed for each subject. The excursion of the Achilles tendon about the ankle was calculated from plantar flexion angle and tendon moment arm [5]. Work-loop plots [6] were generated to evaluate the plantar flexor force and work generated about the ankle (Fig. 1). Slack length was identified as the onset of force generation. Effective stiffness was taken as the slope of the linear portion of the force-length curve during stance. Muscle-tendon work represents the area within the force versus length loop. Forces and work were normalized to body mass. Effective stiffness was normalized by the product of body weight and moment arm at a neutral ankle angle.

RESULTS

Plantar flexor work loops patterns in eqinus gait were substantially different than those seen in typically developing gait. Notably, equinus gait was associated with substantially lower plantar flexor slack lengths, reduced stiffness and diminished work production during pushoff (Table 1). These differences cannot be attributed to walking speed, with both groups exhibiting comparable normalized speeds [3]: 0.40 (0.02) for children with CP, 0.35 (0.04) for controls.

Table 1. Work-loop metrics (mean (SD)) from children with CP walking in equinus compared to controls.

Subjects	Effective stiffness (normalized)	Effective stiffness R^2	Net work (J/kg)	Soleus slack length (mm)
Controls (n=7)	17.58 (5.87)	0.96 (0.04)	0.25 (0.18)	-3.71 (1.60)
Children w/ CP (n=2)	12.13 (1.47)	0.94 (0.01)	-0.02 (0.10)	-5.69 (1.92)

DISCUSSION

This study utilized shear wave tensiometry, kinematics, and EMG data to objectively assess the plantar flexor force and work patterns underlying equinus gait. These analyses revealed substantial and profound differences in plantarflexor behavior. Most notably, the functional plantar flexor lengths were shorter and exhibited strut-like behavior, producing little to no work during pushoff [7]. In contrast, the plantar flexors typically act a motor, generating a substantial amount of positive work. These exciting data demonstrate the potential for using shear wave tensiometers to objectively evaluate muscle contributions to pathological gait, which in turn could enhance treatment planning and outcome evaluations.

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UNDERSTANDING DAILY WALKING ACTIVITY IN THOSE WITH CHRONIC STROKE

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INTRODUCTION

Numerous factors have been shown to be associated with post stroke walking activity, including measures of physical capacity [1], physical health [2], biopsychosocial factors [3], and social and physical environmental factors [4]. However, current models leave much of the variance in post stroke walking activity unexplained. Thus, the purpose of this study was to understand what variables are important in distinguishing subgroups of stroke survivors with respect to their daily walking behavior.

CLINICAL SIGNIFICANCE

Understanding important factors that distinguish walking behavior amongst subgroups of stroke survivors will help clinicians target these factors through intervention.

METHODS

Data for this study was obtained from a four-site randomized controlled trial (NCT02835313) [5] in which daily walking activity was recorded during all waking hours (except bathing) using the FitBit™ that was placed on the participant's non-paretic ankle [6]. In total, our sample included 249 individuals at least 6 months post stroke with a mean age of 63 (SD 11.94), 120 male participants (48.2%), and mean time since initial stroke of 47.5 months (SD 60.2). The average self-selected gait speed of our sample was 0.71 m/s (SD 0.21).

Mixture modeling analysis was used to understand if (and how) subgroups of stroke survivors differ with respect to daily walking behavior [7]. Specific variables of interest included representation of important domains that have been shown to be related to walking activity post stroke and included: Six Minute Walk test, self-selected gait speed, Montreal Cognitive Assessment (MoCA), LDL cholesterol levels, Charlson Comorbidity Index (age-adjusted), body mass index (BMI), Activities-Specific Balance Confidence Scale (ABC), Patient Health Questionnaire (PHQ), living situation, marital status, Area Deprivation Index (ADI), and Walk Score. Average steps per day (SPD) was entered into the model as a distal outcome to understand how the latent classes differed in daily walking activity.

RESULTS

After testing models of various class sizes, a 3-class model was selected based on model selection criteria, including model fit statistics [7]. As there were statistically significant differences in average SPD between each of the 3 classes, the classes were subsequently labeled Low SPD, Medium SPD, and High SPD. Table 1 displays the statistically significant differences in average SPD between each of the classes as well as the number of individuals within each class.

Table 1: Significant Differences in Average SPD between Classes (mean± SD).

	Low SPD (n = 81)	Medium SPD (n= 60)	High SPD (n= 108)
Average SPD*	2567.17 ± 1852.67	4059.36 ± 2078.51	6003.08 ± 2904.14

*All comparisons significant at P<.05

There were no significant differences between the 3 classes with regards to the MoCA, LDL cholesterol levels, Charlson Comorbidity Index, BMI, PHQ, living situation and marital status. However, statistically significant differences were observed between the 3 classes with regards to the Six Minute Walk test, self-selected gait speed, ABC, ADI, and Walk Score (Table 2).

Table 2: Significant Differences between Classes for Variables of Interest (mean± SD).

Variable	Low SPD	Medium SPD	High SPD
Six Minute Walk (m)*	146.31 ± 42.28	277.40 ± 55.39	402.60 ± 65.55
Self-selected Gait Speed (m/s)*	0.42 ± 0.08	0.68 ± 0.09	0.89 ± 0.08
ABC*	64.23 ± 20.04	74.21 ± 17.21	83.0 ± 15.20
Area Deprivation Index^{+}	43.98 ± 27.78	38.91 ± 23.14	31.58 ± 21.46
Walk Score^{++}	33.53 ± 28.34	42.72 ± 28.60	24.97 ± 24.49

*All comparisons significant at P<.05
$^{+}$Statistically significant differences observed between Low SPD vs. High SPD and Medium SPD vs. High SPD at P<.05
$^{++}$Statistically significant differences observed between Medium SPD vs. High SPD at P<.05

DISCUSSION

Results from this study demonstrate that the most important factors that distinguished the 3 subgroups of stroke survivors with different daily walking activity included measures of physical capacity, biopsychosocial factors, and environmental factors. Results from this study aide in our understanding of important factors associated with post-stroke physical activity levels and may help in the design of targeted interventions aimed at improving physical activity after stroke.

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A Long-term Comparison of Lateral Column Lengthening and Medial Calcaneal Sliding Osteotomy for Correction of Pes Planovalgus Deformity in Cerebral Palsy

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INTRODUCTION

Pes planovalgus (PPV) is the most common foot deformity among individuals with bilateral cerebral palsy (CP) and the likelihood of it developing increases with age [1]. Surgery is indicated if the deformity is not reducible with more conservative methods. The aim of surgery is to reduce the subluxed talar head and restore the forefoot and midfoot alignment to achieve a stable plantigrade foot with improved functional ability during walking and standing [2]. A variety of surgical options are available to treat PPV including tendon transfers, osteotomy, arthroereisis and arthrodesis. Calcaneal osteotomies are powerful reliable surgical procedures also used for the correction of PPV foot deformity. Multiple osteotomy techniques are currently used to correct segmental malalignment associated with PPV. Lateral calcaneal lengthening osteotomy (LCL) and the translational medial calcaneal sliding osteotomy (MCSO) are two techniques used to correct hindfoot alignment. Unfortunately, the decision over which calcaneal osteotomy is to be performed is often based on physician preference and no current indications exist for one procedure over the other. Furthermore, reports of post-operative follow-up have not evaluated long-term surgical outcomes once these individuals have transitioned to adulthood and entered the workforce. Therefore, the purpose of the current work was to evaluate the longer-term effectiveness of surgery for PPV and identify the most effective technique that minimizes gait impairment for adults with CP.

CLINICAL SIGNIFICANCE

Long-term assessments of surgical outcomes for children with CP are important because continued growth and weight gain throughout puberty can affect short-term improvements when surgery is performed at a younger age. Continued research and development for surgical techniques and outcomes are indicated, as calcaneal osteotomy will foreseeably remain an important reconstructive option in the correction of foot deformity.

METHODS

Eighteen individuals with bilateral CP treated with either LCL or MCSO for PPV during childhood (13 males, Average age: 24.3 ± 7.9 yrs, Average years since surgery: 10.9 ± 8.4 yrs, GMFCS I: 1, GMFCS II: 7, GMFCS III: 10) participated in a single instrumented gait analysis for the current, IRB approved, prospective study. Additionally, 11 healthy adults with a rectus foot type (3 males, Average age: 26.0 ± 3.5) participated as a Control Group for comparison. One foot was analyzed per subject based on surgical site and/or dominance. The Milwaukee Foot Model (MFM), a segmental foot model, provided kinematic data for four foot and ankle segments [3]. Three representative trials were selected for analysis. Average kinematic curves were calculated and between group comparisons were assessed using the method of locally weighted regression smoothing with alpha-adjusted serial Welsch t-tests (LAAST) [4]. Comparisons were made among each surgical group and the Control Group.

RESULTS

In the coronal plane of the hindfoot, the MCSO Group more closely resembled the Control Group. The LCL Group showed residual hindfoot eversion with compenstory forefoot varus. In the transverse plane of the forefoot, the LCL Group presented with less forefoot abduction. Both groups showed a combination of decreased hindfoot dorsiflexion and lack of forefoot plantarflexion.

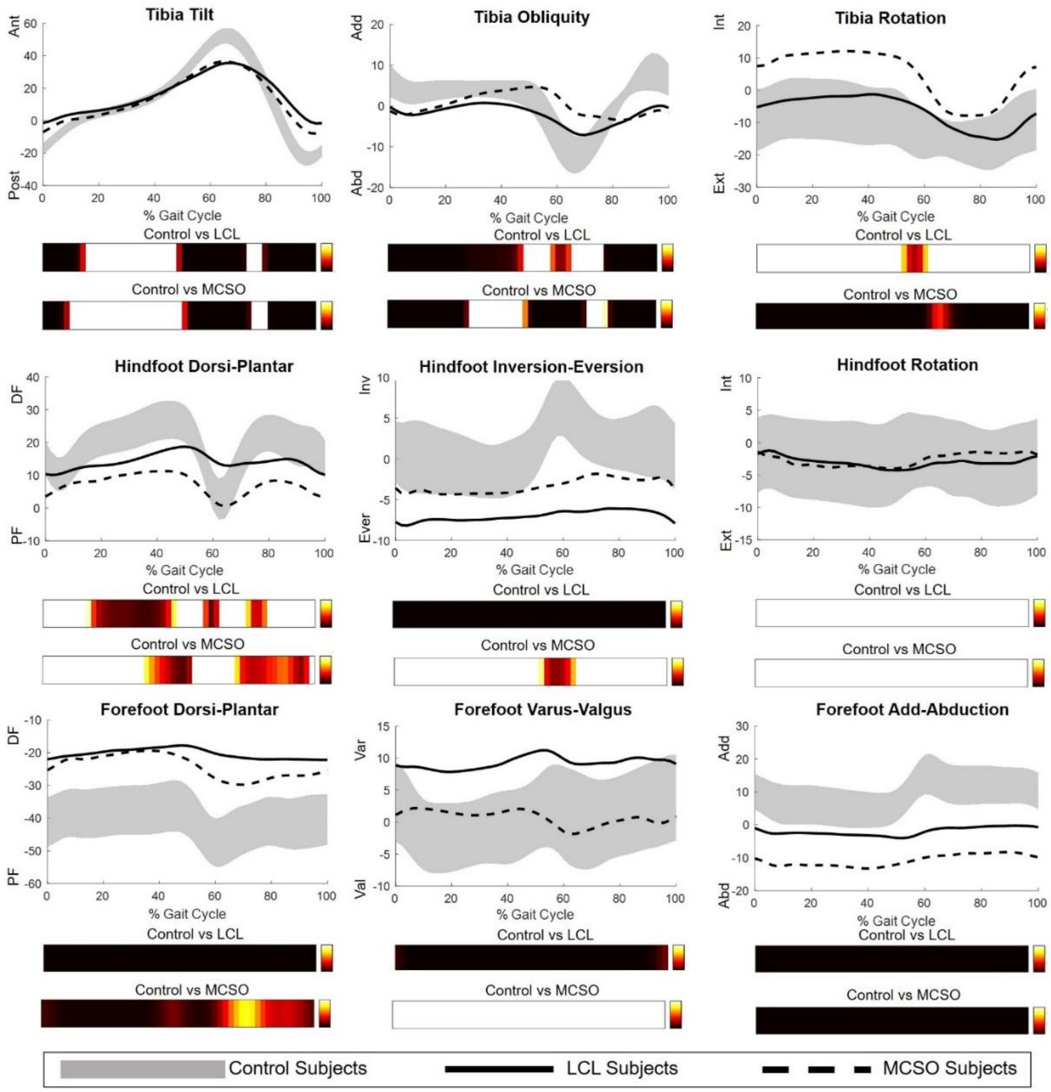

Figure 1: Average multi-segment foot and ankle kinematics for the Control, LCL, and MCSO Groups. Heat maps with adjusted alphas indicate differences throughout the gait cycle between groups.

DISCUSSION

These preliminary long-term results showed that the MCSO Group more closely resembled the Control Group for the coronal plane alignment. Consistent with the goals of a calcaneal lengthening osteotomy, the LCL Group demonstrated less long-term forefoot abduction. Future studies should consider including pre-operative data to truly evaluate surgical correction.

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

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SPATIOTEMPORAL GAIT PARAMETERS OF INDIVIDUALS POST-STROKE WITH USER-DRIVEN TREADMILL CONTROL

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INTRODUCTION

Each year, 795,000 people experience a stroke, which is a leading cause of disability, including impaired gait function and balance [1]. Post-stroke participants have different gait parameters overground compared to on a treadmill [2], as well as compared to healthy controls [3]. Although it is known that user-driven treadmill (UDTM) control allows healthy subjects to select a faster comfortable walking speed [4], it is unknown how the UDTM control affects spatiotemporal parameters in a post-stroke population. **This study examined the effects of UDTM control on post-stroke gait measures to compare the UDTM to a fixed-speed treadmill (FSTM) at matched speeds.**

CLINICAL SIGNIFICANCE

This research demonstrates the potential benefits of a user-driven treadmill control on post-stroke spatiotemporal gait parameters. The UDTM maintains the benefits of FSTM control, specifically increased repetitions in a controlled environment [5], while also encouraging healthy gait measures to provide a more efficient and realistic training environment. Rehabilitation and gait training plans may be altered to use user-driven treadmill control to improve post-stroke gait rehabilitation.

METHODS

Eighteen participants with chronic post-stroke hemiparesis were included in analysis (10M, 8F; 62±11.9 years; 40±30 months post-stroke; 1.73±0.12 m; 84.9±12.9 kg). Subjects walked on an instrumented split-belt treadmill (Bertec Corp., Worthington, OH, USA) while motion capture data was collected (Motion Analysis Corp., Santa Rosa, CA, USA).

Subjects performed a 10-m overground walking task and walked on the fixed-speed treadmill (FSTM) and UDTM at their self-selected (SS) and fast walking speeds. To find the subject's SS and fast walking speeds, the treadmill was adjusted in 0.05 m/s increments from the participant's 10-m speed until the participant reached their preferred speed. Subjects performed six walking trials, each for one minute. The first four trials were randomized:

1. FSTM at SS Speed
2. FSTM at Fast Speed
3. UDTM at SS Speed
4. UDTM at Fast Speed

Then, the subjects performed two speed-matching conditions in a random order:

5. FSTM at UDTM SS
6. FSTM at UDTM Fast

The primary variables of interest were step width, step length, and step time, which were averaged over the one-minute trial for each walking condition. One-way repeated measures ANOVAs were performed to compare the data between the six conditions, blocking for subject, with Tukey post-hoc analysis performed when initial significance was indicated by the ANOVA (α=0.05).

RESULTS

Step width was significantly narrower on the UDTM than the FSTM, even at matched speeds (Figure 1A). There were no significant differences in step width between the four FSTM conditions, showing no effect of speed on step width.

Paretic and non-paretic step length were significantly different between the SS and Fast conditions but did not differ between FSTM or UDTM control (Figure 1B). Additionally, paretic and non-paretic step time differed with walking speed, but not treadmill control (Figure 1C).

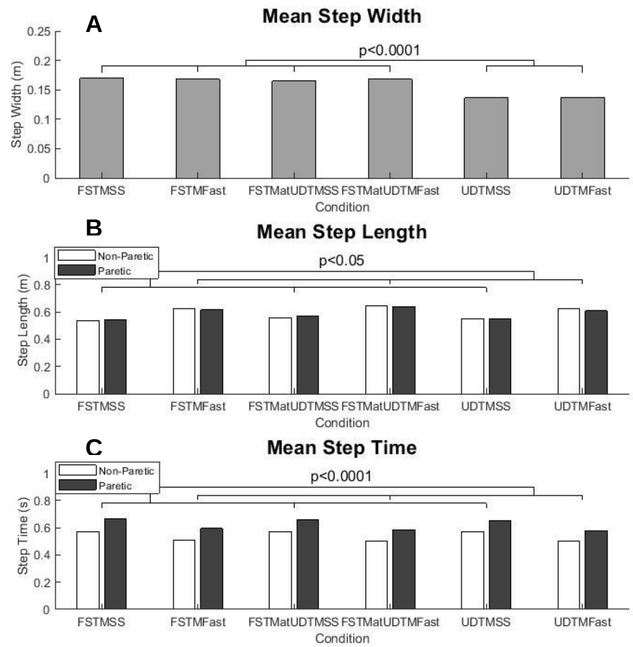

Figure 1: Graphs showing (A) mean step width, (B) mean step length, and (C) mean step time for each experimental condition.

DISCUSSION

The FSTM step width is similar to the values reported by Chen et al. (17.3 cm), while the UDTM value (13.7 cm) shows trends towards the healthy step widths with FSTM (11.5 cm) [3] and overground (12 cm) [6]. Narrower step widths may show an increased sense of stability with the UDTM, which is vital after stroke [1,3,7]. The UDTM may encourage healthy gait mechanics by promoting healthier step widths values.

Especially for the matched speed conditions, step length and time were not expected to differ, since speed is a function of cadence and step length. Although speed does not change in post-stroke individuals with UDTM control [8], the UDTM provides all the benefits of the FSTM while also encouraging improved stability for a more effective training environment.

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DISCLOSURE STATEMENT

The authors have no conflicts of interest to disclose.

WALKING SPEED IS RELATED TO PATIENT REPORTED OUTCOMES OF PHYSICAL FUNCTION IN ADULTS WITH CEREBRAL PALSY

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Introduction

Cerebral Palsy (CP) is the most common motor disability of childhood but presents considerable challenges across the lifespan. A common goal among individuals with CP and their caregivers is to develop and improve walking ability during their growing years and maintain a proficient level of walking into adulthood to maximize mobility and independence[1]. Recent studies also suggest that continuing independent walking throughout adulthood has health benefits beyond mobility, as it promotes general fitness and increased physical activity that can help avoid the negative physiologic consequences of a sedentary lifestyle.

Walking speed is the simplest measure of overall gait performance but has been shown to be a valid, reliable, and sensitive measure appropriate for assessing functional status and overall health in a wide range of populations[2]. However, it has not been widely used as a health indicator in adults with CP. The NIH Patient Reported Outcomes Measurement Information System (PROMIS) is a validated instrument that assesses a patient's physical mental and social health and includes validated subscales in each domain[3]. Finding a positive correlation between these variables can establish the utility of using walking speed as a simple biomarker of physical function in CP that could be easily and regularly monitored. Therefore, the goal of this study is to evaluate the relationship between walking speed and patient reported outcomes of physical function in an adult cohort with CP who had previously been treated at our Center as children.

Clinical Significance

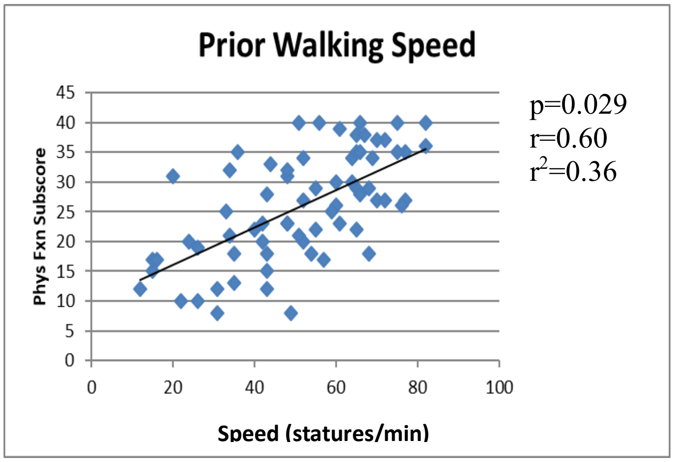
Establishing a relationship between walking speed and more global physical function in adulthood will provide further evidence that walking and health are related in CP. This can influence adult providers to recommend treatments to preserve independent walking in their patients and empower pediatric providers to aggressively improve walking performance before their patients reach skeletal maturity.

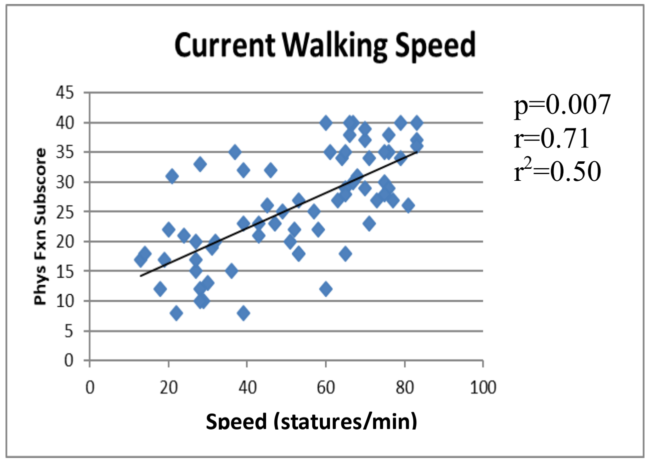
Methods

The current study represents a secondary analysis of data collected as part of the Cerebral Palsy Adult Transition Study (CPAT)[4]. After consent and enrollment, CPAT participants underwent three visits with the research team, completing cognitive tests, surveys, blood work, isometric strength and fatigue tests, the PROMIS-57 and a full instrumented gait analysis (IGA) including temporal-spatial measures. Linear regression was used to assess the association of walking speeds, normalized to height, from each subject's adult IGA and their IGA as children, with the *Physical Function domain sub-score of the PROMIS* (PROMIS-PF). Correlation between childhood and adult walking speeds and PROMIS-PF were assessed using Pearson's correlation coefficients, r, at a significance level of $p<0.05$.

Results

A total of 72 adults with CP (age range: 18.5-48.7 years; mean age[STD]: 25.0[5.3]; GMFCS level I=38.89%, II=40.27%, III=18.06% IV=2.78%; 47.2% Male, 52.8% female, participated in the study. Childhood walking speeds were recorded an average of 8.4[3.4] years earlier than the adult walking speed and PROMIS-PF recordings. Significant correlations were found between childhood *(prior)* walking speed and PROMIS-PF and adult *(current)* walking speed and PROMIS-PF. Pearson r and coefficient of determination are shown in each figure.





Discussion and Conclusion

This study provides evidence that normalized walking speed is positively correlated with a more global physical function measure obtained from a widely used instrument of patient reported outcomes in a cohort of young adults with CP. A strong positive correlation was found between walking speed and the PROMIS-57 physical function sub-score (PROMIS-PF) recorded at the same time. Since the PROMIS-PF is a validated instrument that assesses real-world daily activities of physical function beyond the act of walking, such as ease of accomplishing chores, errands, and physical labor, this confirms that improved walking ability is associated with better overall physical function in this young adult cohort. Furthermore, there is also evidence of good correlation between childhood walking speed and *future* physical performance after skeletal maturity. This is significant as it suggests achieving and maintaining higher levels of walking ability in childhood can lead to improved overall function in daily activities later in life, a finding that has far reaching implications for the treatment of cerebral palsy across the lifespan.

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GAIT PATTERNS OF PATIENTS WITH PROGRESSIVE SUPRANUCLEAR PALSY

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INTRODUCTION

Progressive supranuclear palsy (PSP) is a rare neurodegenerative disease comprised of postural instability, falls, and vertical supranuclear gaze palsy with no proven treatment. These patients often show midbrain atrophy on neuroimaging but definitive diagnosis can only be made on neuropathological exam. Although the classic PSP syndrome presents with clear clinical signs in its later stages, several clinical variants are less distinctive, causing delay in diagnosis of up to 3-4 years after onset of first symptom [1]. Additionally, PSP may be confused clinically with other syndromes in which the clinical features overlap with those of PSP, such as Parkinson's disease, multiple system atrophy, and Alzheimer's disease [2-4]. For this reason PSP may be underdiagnosed and not well represented in previous studies.

Gait disturbance is recognized as one of the core diagnostic features and may present as short, shuffling steps, gait freezing, lurching unsteady gait or spontaneous falls. While temporal-spatial gait characteristics have been previously characterized [5], no comprehensive description of gait in patients with PSP exists. The main goal of the study was to analyze the gait characteristics of patients with PSP.

CLINICAL SIGNIFICANCE

Assessing gait characteristics may help identify distinguishing features of PSP, permit further understanding of affected gait control pathways, distinguish PSP from other causes of parkinsonism, and ultimately contribute to tracking disease progression and development of more effective, disease specific treatment options.

METHODS

The Institutional Review Board approved this study consisting of 10 subjects with PSP (7/10 male, 70.0±6.7 years, BMI: 26.5±4.6 kg/m^2). Subjects walked barefoot and unassisted on a 10 meter walkway for a minimum of three trials. Kinematic data collected at 120 Hz with a 10 camera motion capture system (Raptor 12HS, Motion Analysis Corp, Santa Rosa, CA) and kinetic data collected from five force plates (Optima HPS, AMTI, Watertown, MA) embedded in the walkway were processed in Visual3D (C-Motion, Inc., Germantown, MD). Data for all subjects were compared to a young, healthy adult population collected within the laboratory (10 female, 26.7±5.0 years, BMI: 23.2±2.2 kg/m^2).

Each subject was graded on two clinical scales: the Unified Parkinson's Disease Rating Scale (UPDRS), which evaluates motor symptoms, and the PSP rating scale (PSPRS), which evaluates both motor and non-motor symptoms such as cognition and autonomic function. For both scales, a higher score indicates greater disability. Correlations between gait parameters and the clinical scales were determined using Spearman correlation coefficients. Statistical significance was set to $p \leq 0.05$.

RESULTS

There were significant correlations between UPDRS and PSPRS with velocity (r_s=0.697, *p*=0.025; r_s=0.806, *p*=0.005) and the gait stability ratio (r_s=0.709, *p*=0.022; r_s=0.673, *p*=0.033). The gait of the subjects with the lowest (subjects 1 and 2) and highest (subjects 3 and 4) clinical scores for the UPDRS and PSPRS revealed considerable differences. Distinct differences can be seen within subjects, with subjects with the lowest clinical scores presenting with nearly normal gait patterns and progressively abnormal gait patterns with increasing clinical severity (Figure 1). The subjects with the highest scores, indicating the greatest disability, demonstrated slower walking velocity and reduced sagittal plane ROM compared to subjects with the lowest scores (Table 1). While differences can be appreciated between subjects with PSP, all subjects had abnormal kinematics, kinetics, and temporal-spatial parameters when compared to healthy adults.

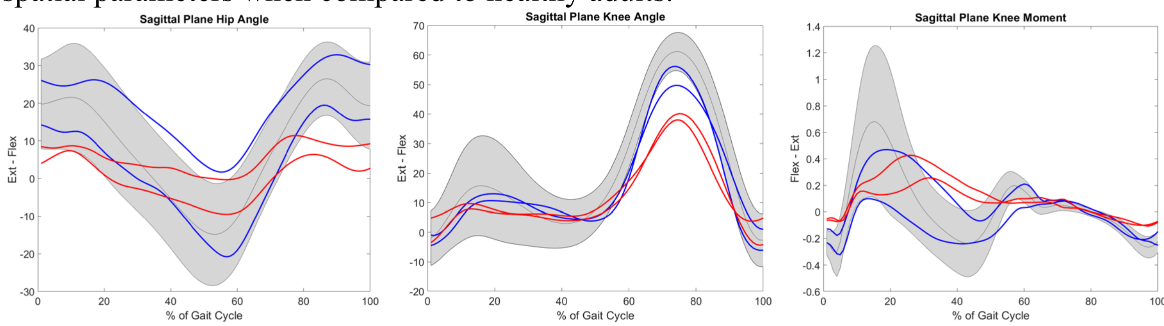

Figure 1: Kinematic and kinetic gait patterns of subjects with PSP. The subjects with the lowest clinical scores (blue) achieve greater range of motion than those with higher scores (red). The subjects with the lowest scores also achieved flexion moments in midstance, where the subjects with higher scores only produced extension.

DISCUSSION

The opportunity to characterize the gait of patients with a rare disease demonstrates that although all individuals may be characterized as having a gait disturbance compared to an unimpaired population, there was a wide range of functional abilities that align closely with clinical ranking scales. With increased disease severity, gait patterns become progressively more abnormal. The significant correlation that gait parameters have with clinical ranking scales makes gait analysis a useful objective tool in evaluating the subtle differences in functional status of this population and provides the ability to track disease progression.

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Table 1: Clinical scores and gait parameters of subjects with PSP

Subject	UPDRS	PSPRS	Velocity (cm/s)	GSR (step/m)	Hip ROM	Knee ROM	Ankle ROM
1	20	31	117.7	1.4	40.3	62.3	23.2
2	36	24	103.3	2.0	34.6	50.9	9.6
3	85	55	54.0	3.4	16.9	44.5	14.3
4	73	58	42.2	3.9	11.7	41.7	14.9

The relationship between pelvic-hip musculature and functional ambulation in patients with Myelomeningocele

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INTRODUCTION

Myelomeningocele (MM) is a neuromuscular disorder that results in bulging of the spinal cord and subsequent muscle paralysis and/or weakness. The lower the level of defect, the more function is preserved. Consequently, patients with defects at the lower lumbar level and below preserve their walking ability. However, our clinical observations suggest that a group of them maintains the ability to walk independently with the use of orthotics (IAs), while another group requires the use of external support, i.e., assistive devices, such as walkers or crutches (NIAs). Currently, to the best of our knowledge, there does not exist any comprehensive literature explaining the underlying cause of this discrepancy. Therefore, the purpose of this study was to investigate if the ambulatory status of patients with lower lumbar MM, IAs compared to NIAs, is related to their muscle strength.

CLINICAL SIGNIFICANCE

Currently, to the best of our knowledge, there does not exist any comprehensive motion analysis studies relating the ambulatory status of patients with lower lumbar level MM to their muscle strength. If this information were known, then more targeted interventions could be identified so that, potentially, NIAs can progress to IAs.

METHODS

This is a retrospective study involving 50 patients with MM who underwent three-dimensional gait analysis (3DGA) in our laboratory as part of their health care plan. For the purposes of this investigation, we focused on their walking pattern characteristics, along with the lower extremity Manual Muscle Test (MMT) portion of their physical exam. Muscle strength from the MMT was translated from the clinical scale [1] to a continuous scale from 1 to 24, where a score of 1 is the equivalent of 0 in the clinical scale, defining the lack of any ability to contract the muscle. A score of 24 in the continuous scale is the equivalent of 5 in the clinical scale, meaning that the patient can move the distal end of the joint it attaches to, through the full range of motion of that joint against maximum manual resistance. Kinematic and temporal-spatial characteristics were determined by the 3DGA involving implementation of a modified Helen-Hayes marker set using specific anatomical landmarks of the segments of the lower body (pelvis, thighs, lower legs, and feet). The averages from each parameter for each group were statistically analyzed using an ANOVA. Statistical significance was achieved at $\alpha \leq 0.05$.

RESULTS

Table 1 shows the relationship between muscle strength for specific muscle groups and ambulatory status. Of the muscle groups investigated, the iliopsoas, sartorius, and gluteus maximus were stronger in the IAs group (p=.0003, p = 0.014, and p=.0002, respectively). Additionally, the group of IAs demonstrated a higher velocity and cadence (p=.02 and p=.005, respectively). Of the kinematic variables assessed, pelvic protraction was greater for IAs (p=.02) and downward pelvic obliquity was greater for the NIAs (p=.02) (Table 2).

Table 1: Statistical output comparing the muscle strength between patients who ambulate independently and those who do not. Muscles tested: adductors (Add); quadriceps (Quad); iliopsoas (Ilio); sartorius (Sar); gluteus maximus (Glut Max); gluteus medius (Glut Med).

Parameter	Add	Quad	Ilio	Sar	Glut Max	Glut Med	Vel	Cad	Stride Length
IA Average ± std. dev.	**18.10** ± 4.99	**20.79** ± 3.36	**20.12** ± 3.58	**18.11** ± 4.23	**10.69** ± 4.37	**7.48** ± 3.31	**0.798** ± 0.03	**0.944** ± 0.02	**0.852** ± 0.03
NIA Average ± std. dev.	**17.45** ± 4.10	**19.71** ± 3.16	**15.25** ± 2.99	**15.00** ± 1.85	**5.92** ± 2.19	**6.88** ± 2.50	**0.643** ± 0.03	**0.761** ± 0.17	**0.833** ± 0.02
p-value	0.7008	0.3661	***0.0003***	***0.0136***	***0.0002***	0.5607	***0.0243***	***0.0049***	0.7504

Table 2: Comparison of the kinematics of pelvic rotation and pelvic obliquity between IAs and NIAs.

Parameter	Pelvic Protraction	Pelvic Retraction	Pelvic Obliquity Upward	Pelvic Obliquity Downward
IA Average ± std. dev.	**18.26** ± 7.41	**-16.70** ± 8.24	**6.74** ± 3.49	**-2.88** ± 4.11
NIA Average ± std. dev.	**13.74** ± 5.54	**-12.54** ± 6.51	**8.44** ± 4.42	**-6.56** ± 4.90
p-value	***0.0229***	0.0652	0.1997	***0.0178***

DISCUSSION

The purpose of this study was to investigate the relationship between ambulatory status of patients with lower lumbar MM who are IAs or NIAs and their respective muscle strength. The stronger hip flexors and extensors seen in the IAs group can provide greater hip stability which can facilitate independent ambulation. Furthermore, the greater velocity of the IAs, which, however, appears to be a function of increased cadence, may be further reflective of the improved ability of the IAs to control their gait compared to the NIAs. The greater excursion range for pelvic rotation in the IAs group may reflect that they achieve their stride length, partly, from pelvic rotation. The assisting devices, on the other hand, may be the limiting factor for the decreased pelvic rotation seen in the NIAs group. The decreased pelvic obliquity in the cohort of IAs may suggest that they have greater pelvic control while ambulating. One limiting factor of this investigation may be the discrepancy in sample sizes between the groups (35 IAs versus 15 NIAs). It is possible, therefore, that there may be some outliers influencing the results.

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ACKNOWLEDGEMENTS

Andrew Moseley-Gholl and Shannon Villegas have been instrumental to this study for the data extraction from patient records.

DISCLOSURE STATEMENT

The authors have nothing to disclose with respect to this investigation.

Progression of Hip Instability in Children with Spinal Muscular Atrophy

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INTRODUCTION

Spinal Muscular Atrophy (SMA) is a genetic neurodegenerative disease with an autosomal recessive inheritance pattern and is considered to be one of the most common genetic causes of death in childhood [1]. While hip subluxation and dislocation are often seen in children with SMA, medical complexity and reduced expected life span of these children has typically resulted in absence of monitoring and subsequent intervention for pathologic hips. However, recent advancements in therapies have resulted in prolonged life expectancy and increased function in children with SMA, [2] thus making orthopedic management an important consideration.

CLINICAL SIGNIFICANCE

Since the rates and pattern of hip migration and acetabular morphology in SMA have not been well described, this study examines the progression of hip instability across all types of SMA in a pediatric population.

METHODS

Following institutional review board approval, a retrospective chart review was performed on children diagnosed with SMA before the age of 25 who were treated at our institution. All x-rays taken before the age of 18 years containing adequate projections of the pelvis were measured for Reimer's migration index (MI), acetabular depth ratio (ADR), and acetabular index (AI). A linear mixed effects model was fit to serial MI measures of individual hips with fixed effects consisting of SMA type, age at x-ray, and their interaction. ADR and AI measures were similarly modeled following conversion of raw values to z-scores based on the model developed by Novais et al [3]. Slope indicated rate of measure change as a function of age.

RESULTS

Forty-five children (22 males) with SMA types 1-3 were included. Six children were classified as type 1, twenty-five were type 2, and fourteen were type 3. The interaction of age by SMA type was statistically significant ($p=0.01$), indicating a difference in the rate of hip subluxation between the three SMA types as measured by MI. By age 4, MI values were different from one another across all three groups ($p<0.01$) (see Fig. 1). When dichotomized for MI greater than vs. less than 30%, it was found that all type 1 children had subluxated hips by age 4 (see Fig. 2). ADR decreased with age across all SMA types. The slopes of ADR regression lines were negative and statistically significant between the three groups (p=0.002). AI values were higher for all types of SMA, which is opposite that expected in normal hips.

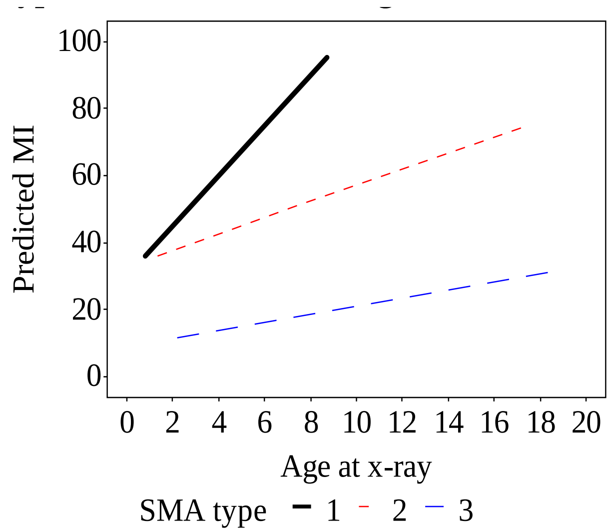

Figure 1. Progression of MI over time across all SMA types.

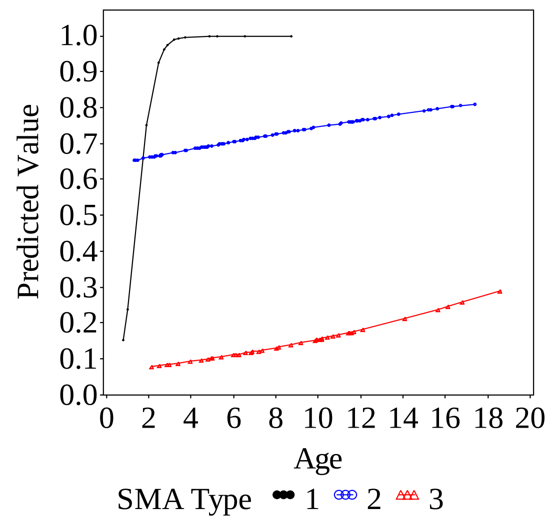

Figure 2. Probability of reaching a MI of >30% across all SMA types.

DISCUSSION

Hip subluxation occurs across all SMA types, with progressively increasing rates and earlier onset in SMA type 1 vs. type 2 vs. type 3. Regression lines of ADR and AI in the SMA subjects compared to those seen in unaffected populations suggest hips in children with SMA do not follow normal adaptive remodeling. As treatments continue to advance and the life expectancies of children increase, there is an increased need to monitor hip instability in pediatric SMA. Future research with standardized imaging protocol is needed to further characterize the natural morphology of hip instability, as well as how this is affected by currently available and developing gene therapies or other treatments, in order to help guide modern treatment decisions such as optimal surgical timing and technique.

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DISCLOSURE STATEMENT

None of the authors have any conflicts of interest to disclose.

Surgical Correction of Bilateral Pes Planovalgus in a Child with Diplegic Cerebral Palsy: A Case Study

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INTRODUCTION

Pes planovalgus (PPV) is the most common foot deformity among patients with bilateral cerebral palsy (CP) characterized by a shortening of the lateral column and a relative lengthening of the medial column. Clinical findings associated with PPV include forefoot valgus and external rotation, and midfoot pronation, as well as hindfoot valgus and equinus. Prolonged PPV may lead to several painful, disabling deformities, including hallux abductovalgus, plantar keratosis, metatarsalgia, and hammer digit syndrome. Surgical treatment of PPV is indicated for patients whose deformity is severe or nonreducible through conservative methods, and who cannot tolerate orthoses. Two current surgical options to correct PPV include calcaneal lateral column lengthening (LCL) osteotomy and subtalar arthroereisis (STA) Peg procedure. The LCL osteotomy translates the navicular bone medially, repositions the talus back over the calcaneus, and corrects the midfoot/forefoot valgus [1]. The STA Peg procedure consists of placing a securely fixed polyethylene endoprosthesis on the dorsal surface of the calcaneus to limit talar movement with respect to the calcaneus [2]. Although these procedures are typically performed in isolation, a combination of the two may be indicated to achieve their individual goals. Therefore, the purpose of the current case report is to describe the successful surgical correction of a child with PPV secondary to CP with a combination of LCL and STA Peg procedures.

PATIENT HISTORY

An 11-year-old female child presented with PPV secondary to diplegic CP, GMFCS level III. She had no prior history of orthopedic surgery. Previous orthotic interventions were unsuccessful in improving her gait pattern, and she used a reverse posture walker prior to surgery.

CLINICAL DATA

The patient presented with a crouch gait and classical PPV including a hindfoot valgus with forefoot varus, reduced forefoot plantar flexion, and forefoot abduction [3]. Physical examination: passive ankle dorsiflexion with knee extended (R1/R2) Left: 0°/5°, Right: 0°/10°; bilateral transmalleolar axes: 23° external; hindfoot eversion: 15°, inversion 0°.

MOTION DATA

The patient underwent an instrumental gait analysis that included multi-segment foot and ankle kinematics using the Milwaukee Foot Model (MFM). Data was collected pre-operatively (age =11.4 years), 1.1 years post-operatively, and 2.6 years post-operatively. Kinematics of interest included sagittal plane ankle, frontal plane hindfoot, frontal plane forefoot, and tibial rotation.

TREATMENT DECISIONS AND INDICATIONS

The patient underwent orthopedic surgical intervention to address her crouch gait that also included bilateral STA peg insertion and bilateral LCL. Post-operatively, she was placed in bilateral knee immobilizers and short leg casts and was non-weightbearing for 4 weeks. She was then provided with bilateral solid AFOs and received intensive physical therapy which included stretching, strengthening, balance and gait training.

OUTCOME

Comparison of the pre-operative and post-operative conditions showed kinematic gait improvements including a decrease in peak ankle dorsiflexion throughout stance from 25° to 8° on the left and from 15° to 8° on the right ankle 2.6 years post-operatively (Figure 1). Additionally, hindfoot eversion improved significantly bilaterally, showing a slightly marked eversion in the left foot and prominent inversion in the right foot. Similarly, the patient's forefoot varus drastically improved bilaterally, showing normal range throughout the gait cycle on the left foot and near-valgus kinematics throughout gait on the right. With these changes also came extreme external tibial rotation on the right LE and near corrected external tibial rotation on the left.

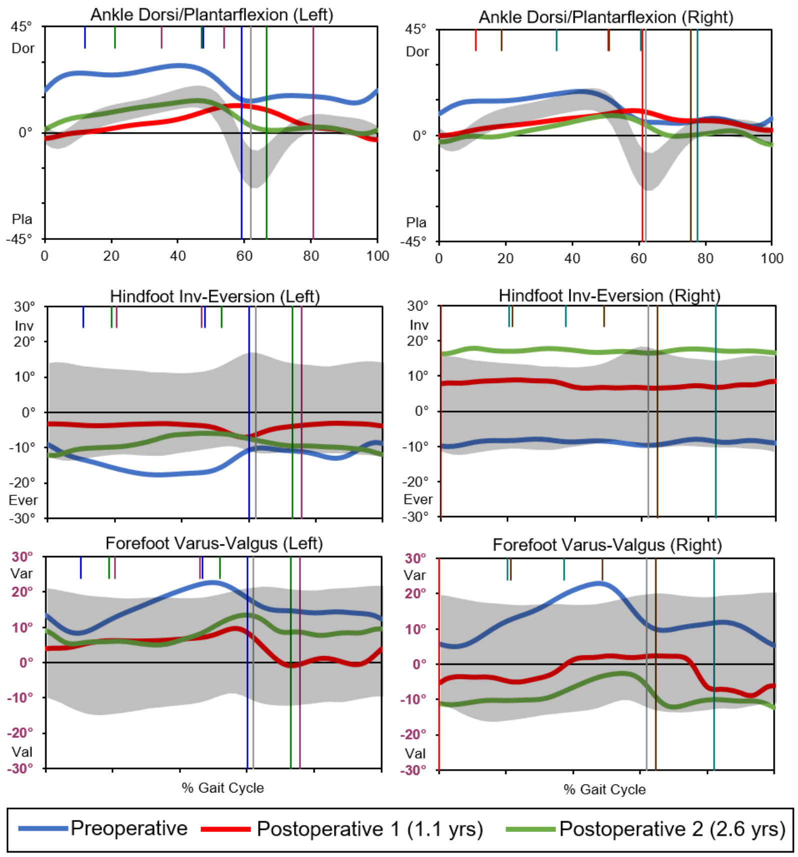

Figure 1: Sagittal ankle and coronal hindfoot/forefoot kinematics at each visit.

SUMMARY

An 11-year-old girl with spastic diplegic CP presenting to the clinic with PPV was indicated for orthopedic surgical intervention due to her persistent dorsiflexion through gait and failure of conservative treatments to improve her condition. Following bilateral STA peg insertion and bilateral lateral column lengthening, the patient showed prolonged improved dorsiflexion bilaterally throughout her gait cycle. In addition to this, she showed marked improvements in her previous hindfoot eversion and forefoot valgus bilaterally throughout her gait cycle. Further assessment is needed to determine the correlation of gait improvement outcomes to the patient's comorbid crouch gait stance and associated surgical correction.

DISCLOSURE STATEMENT

None of the authors have conflicts of interest to disclose.

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COMPARISON OF BETWEEN CHARCOT-MARIE-TOOTH TYPE 1 AND 2 IN TERMS OF LOWER EXTREMITY WORK

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INTRODUCTION: Charcot-Marie-Tooth (CMT) is the most common inherited peripheral neuropathy of childhood that affects 1 in 2,500 people in the USA [1]. The resulting length dependent neuropathy leads to distal muscle weakness and foot deformities that result in gait deviations such as reduced ankle plantar flexor moments and peak powers in comparison to typically developing (TD) peers [2]. Peak moments and powers, however, may not tell the full story related to weakness in the lower extremities that may impact gait during the full stance phase. The computation of mechanical work reflects the production or absorption of mechanical energy at a joint and may be more informative in understanding the implications of weakness which may differ depending on CMT type and disease progression. The goal of this study was to explore the differences in work in stance between the most common types of CMT (CMT1 and 2), [3] in the context of TD reference data.

CLINICAL SIGNIFICANCE: Establishing measures that differentiate between CMT type can assist in understanding the differences in presentation between types, disease progression, appropriate treatment to support gait issues and treatment outcomes all of which are important for pending new therapies.

METHODS: A total of 59 patients with various types of CMT completed instrumented 3D gait analysis following a standardized protocol [4] during barefoot walking as part of a natural history and surgical outcomes study. From this study cohort, patients with CMT1 (29 subjects; 9 females) and CMT2 (10 patients; 6 females) were analyzed (Table 1). For patients with multiple gait analyses, the first was included as test this was likely at the time when gait problems were starting to manifest. Temporal stride parameters and gait kinematic and kinetic measures were extracted using custom MATLAB code (MATLAB R2012b, The MathWorks, Natick, MA, USA,2012) for the patient and TD reference data collected in the same facility. Work was calculated as the numerical integration of the sagittal power curve for each joint. Comparison between CMT groups was performed using a Student's T-Test, alpha=0.05.

RESULTS: The comparison between CMT1 and 2 for selected variables can be found in Table 1. There was a significant difference between groups for age, height and body mass which were all greater in the CMT1 group. There was also a significantly greater walking velocity in CMT1. However, when normalized to leg length, the walking velocity showed no differences between groups. The CMT2 group was found to have significantly less peak ankle plantar flexor moment and power versus CMT1. This was consistent with the significantly reduced positive work at the ankle in CMT2 versus CMT1. The CMT2 group however, showed significantly greater hip positive work than CMT1.

DISCUSSION: The findings in this study show that mechanical work measures can differentiate between CMT1 and 2 and are also different than TD. Consistent with the finding of increased plantar flexor weakness in CMT2 as shown by significantly decreased peak ankle plantar flexor moment and power in terminal stance there was significantly less positive work done by the ankle in CMT2 in comparison to CMT1. The positive work total for CMT2 was about ½ of that found in CMT1 and perhaps better reflects the implication of plantar flexor weakness across the majority of stance phase that is not captured in a peak moment or power measure. The CMT2 group also had significantly greater hip positive work in stance than CMT1. This may reflect the increased demands on the hip extensors in CMT2 to compensate for ankle plantar flexor weakness.

	CMT1 Group	**CMT2 Group**	*p value*	**Typically Developing**
Demographics				
Age (years)	12.6 ± 3.0	9.3 ± 4.4	*0.005**	15.6 ± 7.3
Height (m)	1.53 ± 0.19	1.30 ± 0.19	*0.000**	1.52 ± 0.33
Body Mass (kg)	50.4 ± 19.1	33.3 ± 12.2	*0.000**	50.0 ± 22.0
Temporal Stride				
Walking velocity (m/sec)	1.15 ± 0.18	0.95 ± 0.36	*0.024**	1.23 ± 0.40
Walking velocity/leg length	0.145 ± 0.028	0.143 ± 0.052	*0.869*	0.159 ± 0.055
Sagittal Kinematics				
Peak Dorsiflexion in Mid 1/3 Swing (°)	-3 ± 5.9	-5 ± 7.1	*0.206*	1 ± 3.0
Peak Dorsiflexion in Stance (°)	12 ± 5.6	17 ± 10.5	*0.067*	13 ± 2.9
Peak Knee Flexion Swing (°)	58 ± 7.1	63 ± 13.2	*0.150*	59 ± 5.3
Peak Hip Flexion Swing (°)	35 ± 7.7	50 ± 9.0	*0.000**	32 ± 5.9
Sagittal Kinetics				
Peak Plantar Flexor Moment ST (Nm/kg)	1.18 ± 0.28	0.88 ± 0.31	*0.001**	1.37 ± 0.31
Peak Ankle Power Generation TS (W/kg)	2.91 ± 0.98	1.96 ± 1.20	*0.007**	4.13 ± 1.13
Peak Hip Power Generation PSW-ISW (W/kg)	1.12 ± 0.40	1.41 ± 0.73	*0.125*	1.33 ± 0.50
Sagittal Work				
Ankle Positive Work ST (J/kg)	0.195 ± 0.080	0.101 ± 0.075	*0.000**	0.281± 0.077
Ankle Negative Work ST (J/kg)	-0.149 ± 0.066	-0.172 ± 0.054	*0.155*	-0.119 ± 0.052
Knee Positive Work ST (J/kg)	0.100 ± 0.058	0.132 ± 0.091	*0.186*	0.094 ± 0.038
Knee Negative Work ST (J/kg)	-0.087 ± 0.051	-0.070 ± 0.040	*0.160*	-0.092 ± 0.038
Hip Positive Work ST (J/kg)	0.126 ± 0.058	0.191 ± 0.102	*0.022**	0.112 ± 0.047
Hip Negative Work ST (J/kg)	-0.122 ± 0.070	-0.102 ± 0.064	*0.288*	-0.131 ± 0.060

Table 1. Comparison of mean (±1 S.D.) outcome measures between CMT1 and CMT2 groups, (* indicates significant difference, $p < 0.05$,). Typically developing values provided for reference only. Stance (ST), Terminal Stance (TS), Toe off (TO), pre-swing (PSW), initial (ISW)

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DISCLOSURE STATEMENT

The authors have no conflicts of interest to disclose.

EVALUATION OF PLANTAR PRESSURE AUTOMATED MASKING AND TEMPORAL SPATIAL GAIT PARAMETERS IN CHILDREN WITH SPINA BIFIDA

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Introduction

Plantar pressure (PP) measurements capture differences in foot pressure distribution and are typically segmented into different regions by manually applying a mask. A key part of these masks is the foot progression angle (FPA), which bisects them longitudinally. Temporal spatial gait parameters are used clinically to determine overall gait performance, and both PP and temporal spatial measures can be used to track disease progression over time. Motion capture systems serve as the current best-practice for FPA and temporal-spatial parameters, although use of a plantar pressure mat has advantages such as ease of access and speed of data processing, in addition to acquiring PP distributions unavailable from motion capture. This study aims to compare the accuracy of FPA and temporal spatial data generated from a plantar pressure mat (Novel emed XL, Munich, Germany) to that of a 3D optical motion capture system (Vicon Vantage, Oxford, UK) in a group of children with spina bifida.

Clinical Significance

Previously, automated plantar pressure masks and temporal spatial parameters obtained from a plantar pressure mat were tested in subjects without foot pathology.[1,2] Validation of these techniques in a patient population with atypical foot presentation is necessary to evaluate the automated masking technique in subjects expected to have changes in PP over time.

Methods

Thirty-five participants (mean age 10.7 years, SD 4.4 years) with a diagnosis of spina bifida were recruited from the Spinal Defects Clinic at Children's Hospital Colorado. The participants walked in a motion analysis lab wearing reflective markers on their feet in a fashion similar to the Oxford Foot Model.[3] Optical motion capture cameras tracked 3D locations of the markers while a plantar pressure mat simultaneously recorded foot pressures. Five trials with at least two steps on the mat were collected for each participant as they passed through the motion capture volume and over the PP mat. Marker trajectories were labelled, gap-filled, and filtered in Vicon Nexus and then imported into MATLAB (Mathworks, Natick, MA). Motion capture of steps recorded on the PP mat were matched to the corresponding PP data.

Raw pressure data were imported into MATLAB where foot pressures were automatically identified and centers of pressure (CoPs) were calculated for each step. All viable steps were masked using one manual and five automated techniques. (Image Processing, Heel-Centroid, Full, CoP, 66% CoP, CoP Inter-Peak) These masks were previously applied to subjects without foot pathology.[1]

The timing of foot-floor contact was calculated by identifying steps and using the first and last moments of pressure applied. Foot pressures were automatically identified, and a base heel location was assigned. The heel base point was used to determine step width and length.

Results/Demonstration

Cadence (-0.27 (6.76,-7.30) steps/min, *Mean Difference*, (*Limits of Agreement*)) gait speed (-0.01 (0.06,-0.09) m/s), step length (-0.31 (2.09,-3.58) cm), and step width (1.16 (4.73,-2.41) cm) each had low mean differences between systems and were comparable to our previous study.[2] The limits of agreement indicated room for improvement in plantar pressure analysis of step width however, as 4.73 is 43% of the average motion capture calculated step width. Manual FPA mean difference is below the average kinematic minimal detectable change of ~4 degrees.[4] All automated techniques produced mean difference above four degrees with large standard deviations.

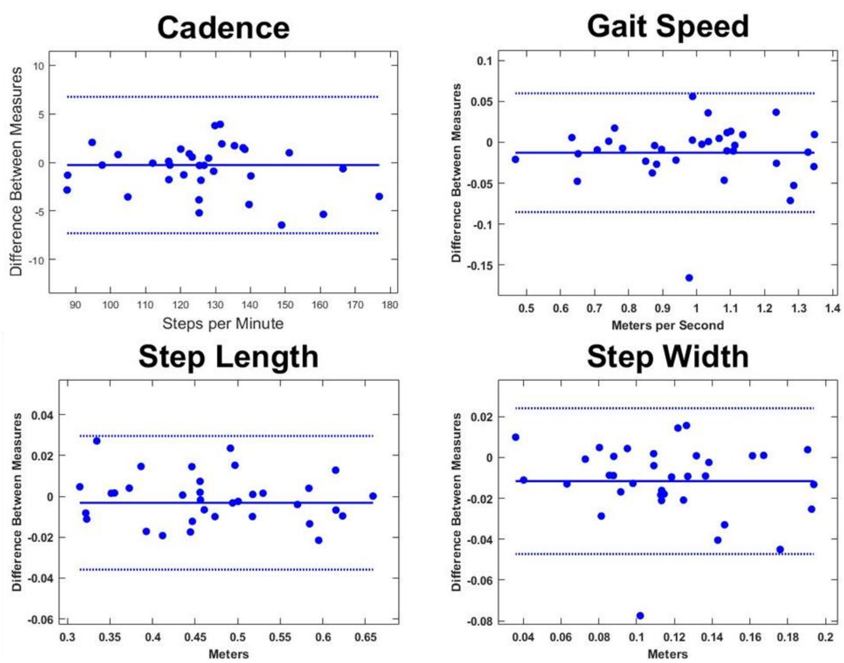

Figure: Bland-Altman plots of the temporal spatial parameters evaluated for the series. The solid line indicates the mean difference between the measurement systems. The dotted lines set the limits of agreement.

Table: Mean difference and standard deviation between optical motion capture and the plantar pressure mat for foot progression angle.

	Participants with Spina Bifida						Participants with Typical Gait				
	Manual	Image Processing	Full CoP	66% CoP	CoP Inter-Peak	Heel-Centroid	Image Processing	Full CoP	66% CoP	CoP Inter-Peak	Heel-Centroid
Mean FPA Difference (Degrees)	3.67	5.69	6.31	8.30	12.15	7.26	2.48	3.30	2.84	3.40	3.40
Standard Deviation (Degrees)	2.45	6.90	3.58	6.17	14.11	7.70	1.72	2.04	2.40	3.19	2.25

Summary

The results show that the plantar pressure mat was able to produce temporal spatial gait parameters with a low mean error in a cohort with spina bifida that is clinically acceptable and similar to the system errors shown in our previous study of normal subjects. FPA mean error was acceptable for manual masking, but unacceptably high for automated masks. This indicates that the complex pressures produced by the spina bifida population will require a different automated masking technique if automation is to replace manual masking.

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Disclosure Statement

The authors have no conflicts of interest to disclose.

FREE VERTICAL MOMENT REFLECTS TURNING ABILITY IN STROKE SURVIVORS

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INTRODUCTION

Turning while walking is a common and essential task to activities of daily living. Stroke survivors were reported to have a higher rate of falls during this movement [1]. A free vertical moment (FVM) of ground reaction is recognized to be the torque about the vertical axis that originating shear forces between the foot and the ground at the foot's center of pressure. FVM was reported to increase during spin turns more than step turns as the foot rotates about the ground in typically developing children [2]. Very little is known about the kinetic characteristics of turning ability of stroke survivors, although their characteristics will be expected to provide useful information to develop a new rehabilitation program for stroke patients to improve their turning ability.

CLINICAL SIGNIFICANCE

The FVM may be one of the kinetic parameters that determine the turning ability of stroke survivors. These findings are expected to be helpful to develop turning gait exercise in stroke rehabilitation.

METHODS

Eight stroke survivors aged 57 ± 7 years (mean ± SD) participated in this study. Participants were asked to start 90-degree turns to the right or left while walking at a comfortable walking speed when the step ipsilateral to the turn direction reached the corner of the walking path, which is defined as the turn step. The steps before and following the turn step was defined as the approach step and the depart step, respectively. Turn phase was defined as the period from the approach step to the depart step. They were not allowed to use any walking assistive devices. Data were collected using an 8-camera motion analysis system and four force plates to compute kinematic and kinetic parameters starting with the approach step, the turn step, and lastly depart step during turning, including a pelvic rotation angle (PRA) and an FVM.

PRA was measured as the angle between a line connecting the reflective markers of both anterior superior iliac spines at the approach step and turn step, respectively.

The FVM was calculated using a previously published equation [3]. Positive values stand for torque acts to resist foot adduction and negative torque acts to resist foot abduction for both sides of the feet. Normalization was carried out using the body weight and the height of each participant. The positive/negative FVM value represents the friction force resisting the abduction/adduction of the foot on each side in this study.

The stride width during turning normalized using the height of each participant.

These data were compared between turns to the affected side and the unaffected side using the paired t-test for parametric data and the Wilcoxon signed-ranks test for nonparametric data. Correlations of the peak FVM of the turn step with the gait spatio-temporal parameters were calculated using Spearman's correlation coefficient.

RESULTS

The peak FVM of the turn step in the external rotation direction was significantly greater during turning to the unaffected side than to the affected side ($8.5\pm3.4\times10^{-3}$ vs. 5.8 ± 4.1 dimensionless$\times10^{-3}$, $p=0.02$). The stride width during turning to the unaffected side was significantly smaller than during turning to the affected side (-0.01 ± 0.04 vs. 0.04 ± 0.03 dimensionless, $p<0.01$). The PRA during the stance phase of the turn step was significantly larger during turning to the unaffected side than to the affected side (74±20 vs. 52±15 degrees, $p=0.02$), while the PRA of approach step was slightly, but not significantly, larger during turning to the affected side than to the unaffected side (7±7 vs. 20±16 degrees, $p=0.11$).

The peak FVM of the turn step in the external rotation direction was negatively correlated with the stride width during turns to the unaffected side (rho=-0.81, $p<0.01$) and to the affected side (rho=-0.86, $p<0.01$). There was no significant correlation between the peak FVM and the PRA of the turn step.

DISCUSSION

We have observed that the FVM showed the biphasic pattern during turning in line with the previous study (e.g., [2, 3, 4]). The stroke survivors had greater peak FVM and PRA of the turn step, and less stride width during turning to the unaffected side than during turning to the affected side and that the peak FVM of the turn step correlated with the stride width during turning. As we did not find a significant correlation between the FVM and the PRA of the turn step, the foot progression angle may affect the FVM because the FVM is considered to reflect the torsional motion of the entire lower limb [4].

Our results suggest that stroke survivors performed turns to the unaffected side mainly using the unaffected lower limb while they performed turns to the affected side using both lower limbs, which means compensating for the declined FVM of the turn step of the affected lower limb with initiating turn with the approach step of the unaffected lower limb. These results can suggest that the FVM of the affected lower limb may be one of the kinetic characteristics that determine the turning ability of stroke survivors.

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DISCLOSURE STATEMENT

The authors declare no financial or personal conflicts of interest.

PROGRESSION OF WALKING VELOCITY IN CHARCOT-MARIE-TOOTH DISEASE

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INTRODUCTION

Youth with Charcot-Marie-Tooth disease (CMT) have progressive increasing weakness that can impact gait function and be measured by walking velocity. It is unclear whether walking velocity continues to decline through childhood and adolescence in comparison to peers and if different types of CMT respond differently in terms of this outcome parameter. The aim of this study was to examine differences in preferred walking velocity as a function of growth and CMT type in a cross-sectional sample of youth with CMT.

CLINICAL SIGNIFICANCE

Since failure to increase walking velocity with age is primarily due to lack of increase in stride length, treatments that enable improved stride length, such as plantar flexor strengthening and bracing, are likely to improve walking speed and associated gait function in youth with CMT.

METHODS

30 youth with CMT type 1 (CMT1) (7 females, mean age at baseline 12.4 years, SD 3.2, range 5.1 to 16.7), 12 youth with CMT type 2 (CMT2) (8 females, mean age 9.3 years, SD 4.2, range 4.3 to 16.2), and 54 controls without disability (mean age 9.6 years, SD 3.4, range 4.7-19.1) were tested in the motion analysis laboratory at a tertiary care children's hospital following standard comprehensive gait analysis protocols [1]. Preferred walking velocity was measured as part of comprehensive gait analysis testing. Some patients were tested more than once resulting in 47 total CMT1 observations and 30 total CMT2 observations between the ages of 4 and 20 years. Changes in walking velocity, cadence, and stride length with age were compared among groups (control, CMT1, CMT2) using linear mixed effect models including a random intercept term to model the repeated measures for some participants.

RESULTS

As expected, walking velocity increased significantly with age in controls (Table 1, Figure 1 top - black line). However, walking velocity in those with CMT demonstrated minimal non-significant change ($p>0.35$). Stride length increased significantly with age in all groups, but to a lesser extent in the CMT groups compared with the controls ($p<0.05$ for group-age interaction terms). Cadence decreased significantly with age at similar rates in all groups ($p>0.57$; Figure 1 bottom right).

DISCUSSION

Youth with CMT lack the normal increase in walking velocity and function with age compared to typically developing peers. This appears to be a more substantial issue and starts earlier for those with CMT2 vs. CMT1. The slower increase in walking velocity resulted from smaller increases in stride length, which is most likely caused by reduced plantar flexor strength [2] and increased ankle lateral instability both common in persons with CMT [3]. Treatments that allow for improved stride length such as plantar flexor strengthening and bracing, which can also improve ankle stability in stance, are likely to improve walking velocity and associated function such as keeping up with peers. These would need to be applied to those with CMT2 at a younger age than those with CMT1.

Table 1: Change in temporal-spatial parameters with age

	Control	CMT1	CMT2
Change in walking velocity (cm/s/year)	2.2 (0.6, 3.7)*	0.8 (-0.9, 2.5)	0.3 (-1.5, 2.1)
Change in stride length (cm/year)	4.5 (3.4, 5.5)*	2.9 (1.8, 4.0)*	2.2 (1.2, 3.2)*
Change in cadence (steps/min/year)	-2.6 (-3.7, -1.6)*	-2.2 (-3.4, -1.1)*	-2.5 (-3.8, -1.2)*

Results are presented as coefficient for age (95% CI). * indicates significant rate of change with p<0.001.

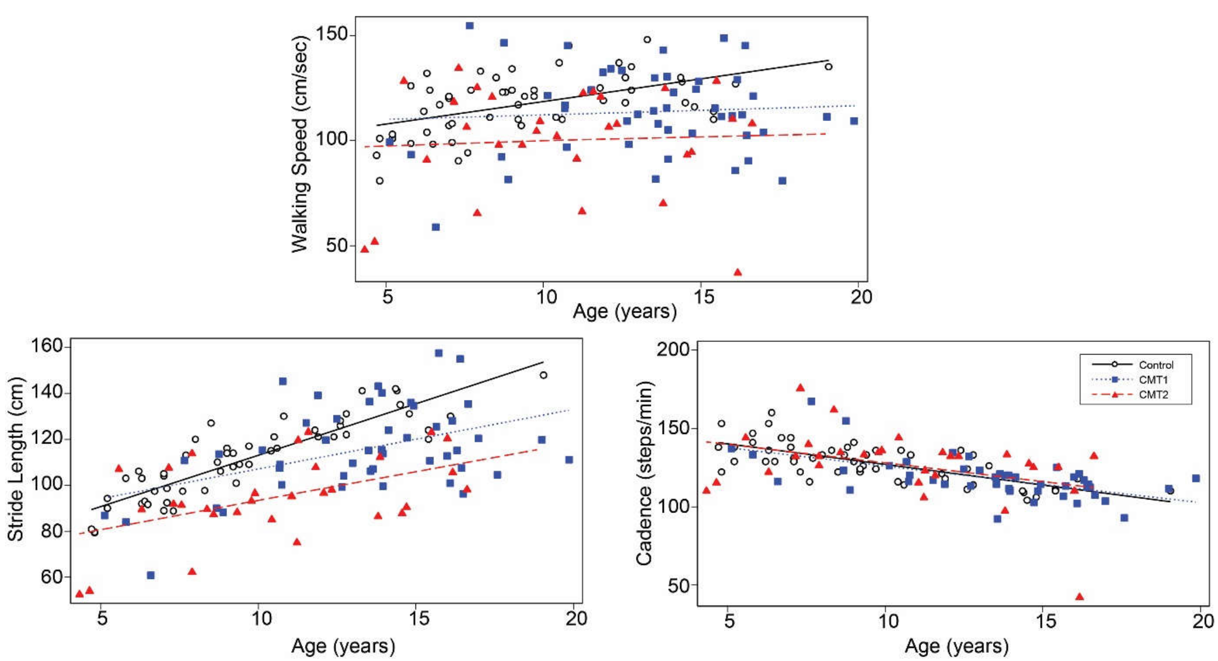

Figure 1: Temporal-spatial parameters as a function of age

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

EVALUATION OF KINEMATIC EMG AS A POSSIBLE BIOMARKER FOR GAIT DECLINE IN YOUTH WITH CMT: A PRELIMINARY ANALYSIS

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INTRODUCTION: Charcot-Marie-Tooth (CMT) is the most common inherited peripheral neuropathy of childhood that affects 1 in 2,500 people in the USA [1]. The resulting length dependent neuropathy leads to distal muscle weakness and associated foot deformities that result in gait deviations in comparison to typically developing (TD) peers [2]. Surface kinematic electromyography (kEMG) can be measured during gait for the key muscles that are impacted by CMT such as the anterior tibialis (which leads to excessive equinus in swing) and the gastrocnemius (which leads to excessive dorsiflexion in stance and reduced plantar flexor power generation). A quantitative analysis of the resulting kEMG signal in the frequency domain could serve as a viable biomarker in CMT to better understand disease onset and progression which are critical for determining the efficacy of emerging new treatments such as gene therapies. The goal of this initial study was to evaluate the changes in kEMG parameters over time and how these changes are associated with gait function at the ankle.

CLINICAL SIGNIFICANCE: If kEMG can show the presence of disease prior to clinical manifestation then it would be an important biomarker for determining disease onset, progression and effects of emerging therapies. The first step is to understand how kEMG relates to gait function as measured using comprehensive 3D gait analysis (3DGA).

METHODS: A cohort of ten patients (3 females, mean age at base line of 11.0 years, SD 2.7, range 5.8 to 14.7 and mean age at follow-up test of 13.6 years, SD 3.1, range 7.7 to 17.0) with a diagnosis of CMT type 1 (CMT1) who had multiple gait analysis tests were selected from a larger study data set. All patients completed 3DGA following a standardized protocol [3] during barefoot walking. This included the collection of surface kEMG from the anterior tibialis (AT) among other muscles during level gait. kEMG data was initially processed using Vicon Nexus (Vicon Motion Systems, Los Angeles, CA) followed by custom Matlab code (Mathworks, Natick, MA). The kEMG signals were high pass filtered (50Hz), then rectified and low pass filtered (10Hz) to create a linear envelope signal. The mean amplitude, root mean square (RMS) and integrated EMG (iEMG) were calculated for the AT during the swing phase. The kEMG signals were also analyzed in the frequency and power spectrum domains. A Fast Fourier Transform of the detrended signals (i.e. the raw signal has a mean of 0 as drift and noise has been removed) was used to convert the time series data into the frequency domain to provide both frequency and power spectrum data. From these domains the median frequency and mean frequency were calculated for the kEMG signals during gait. The preliminary data in this pilot will use descriptive statistics and paired t-tests ($p<0.05$) to begin to understand the possibility of using kEMG as a biomarker for adolescents and youth with CMT.

RESULTS: A comparison of the changes over time in the kEMG measures for the AT only is located in Table 1. The mean frequency data showed a trend towards lower frequencies with minimal change in RMS and iEMG. When looking at individual patients, 12/20 sides showed a decrease and 8/20 sides an increase for iEMG and RMS and mean and median power frequency.

Similarly, 13/20 sides showed a decrease and 7/20 sides and increase for the mean and median frequency. For the kinematic data, 10/20 sides showed an increase in ankle plantar flexion at 98% of the gait cycle, and 11/20 sides showed an increase in ankle plantar flexion in mid swing. The findings were similar on both sides (showing a decrease or increase) for 6/10 patients in the majority of measures.

DISCUSSION: This preliminary study shows that there is a large variation in kEMG measures in patients with CMT1. There is a trend towards lower kEMG frequency measures with time for the AT. The AT iEMG and RMS, however, did not show this trend. When evaluating these outcomes in the context of the associated gait kinematics, there was no change in mean peak ankle dorsiflexion in mid swing and ankle angle just prior to initial contact on average. These findings may suggest that frequency changes in the AT kEMG may manifest earlier than gait findings. However, when evaluating individual patients, about 8/20 sides showed a decline in dorsiflexion which suggests increased impairment at the joint level (ankle dorsiflexor weakness) and a decline in at least one of the kEMG measures. Future studies need to include increased patient numbers and document changes over a greater time frame to further explore the potential of kEMG as a biomarker for adolescents and youth with CMT.

Table 1: Comparison of mean (±1 S.D.) kEMG measures between test 1 and 2 for the anterior tibialis in patients with CMT1. Typically developing values provided for reference only.

	TD	**CMT Type I**		
	n/a	**Test 1**	**Test 2**	**p**
iEMG (v)	812.4	1005 ±482	883 ±592	0.239
RMS (v)	0.89	0.90 ±0.31	0.91 ±0.44	0.457
Mean Frequency (Hz)	137.6	111 ±20	103 ±11	0.074
Median Frequency (Hz)	114.2	95 ±20	86 ±12	0.042
Mean Power Frequency (Hz)	281.1	113 ±24	103 ±15	0.060
Median Power Frequency (Hz)	262.4	95 ±25	83 ±16	0.051
Peak Dorsiflexion mid SW (deg)	1 ±3	-5 ±5	-6 ±4	0.311
Dorsiflexion 98% GC (deg)	0 ±3	-8 ±6	-9 ±4	0.246

iEMG integrated EMG, root mean square (RMS), swing (SW), gait cycle (GC), dorsiflexion = +ve

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DISCLOSURE STATEMENT

The authors have no conflicts of interest to disclose.

EFFECTS OF PROSTHETIC SOCKET DESIGN ON RESIDUAL LIMB MOTION USING DYNAMIC STEREO X-RAY

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INTRODUCTION

Despite the importance of a well-fitting socket for optimum functionality for individuals living with lower limb amputation (LLA), current clinical practice for socket fabrication is often based on unscientific, artisanal methods that lack repeatability. Furthermore, methodology to accurately quantify the three-dimensional (3D) residual limb-socket kinematics is complicated by the multi-factorial nature of the interaction between the residual limb (bone and tissue), the liner, and the socket during dynamic activities. Currently there is little existing data on dynamic, in vivo residual limb-socket kinematics in LLA, as most investigations have been performed referencing non-dynamic testing protocols [1,2]. Highly accurate, dynamic assessments of 3D, in vivo residual limb-socket kinematics are only possible using Dynamic Stereo X-ray (DSX). The purpose of this investigation was to determine the dynamic, in-vivo kinematics between the residual limb and socket for individuals with transfemoral amputation (TFA) using two socket types: a compression release/stabilization (CRS) socket and a traditional (TRAD), encapsulated socket.

CLINICAL SIGNIFICANCE

There remains a fundamental need for accurate, biomechanical evaluations of residual limb-socket kinematics, which can be appropriately translated into evidence-based clinical practice for amputation care, improving overall quality of life for individuals living with TFA.

METHODS

Five individuals living with TFA, all experienced prosthetic users (>6 hours/day), were recruited from the Veteran Affairs New York Harbor Healthcare System (VA NYHHS) and the Providence VA Medical Center (PVAMC). All procedures were approved by the respective Institutional Review Boards. Each subject was fit with a TRAD encapsulated socket and a CRS socket by an ABC accredited and CRS-certified prosthetist [3]. Participants were randomized to start with the TRAD or CRS socket and wore each socket for 4 weeks. Following each 4-week period, participants completed the Trinity Amputations and Prosthetics Experience Scale (TAPES) satisfaction scale, and items related to socket comfort and fit drawn from both the Prosthetic Evaluation Questionnaire (PEQ) and Prosthetic Profile of the Amputee (PPA). Following the 8-week period, subjects were transported to the W.M. Keck Foundation X-Ray Reconstruction of Moving Morphology (XROMM) facility at Brown University where DSX was utilized to record dynamic X-ray sequences simultaneously with optical motion capture (OMC) for 16 trials of treadmill walking at self-selected speed (.36-.80m/s). The DSX system was positioned to capture the residual limb movement at a complementary 60° antero-posterior path during the gait cycle. OMC data was collected using 37 reflective markers on the lower extremities. Following the data collection, subjects received a CT scan for their residual limb and sockets at the PVAMC. All processing of CT and X-ray data was performed with the DSX Suite software (C-Motion, Germantown, MD).

RESULTS

Bone movement was expressed as the excursion of the translations and rotations between TRAD and CRS sockets. The CRS socket had a significantly larger axial translation (mean (SD), 2.03 (0.59) cm) compared to the TRAD socket (mean (SD), 1.55 (0.66) cm; *P*=0.043) (Figure 1). No significant differences were found in the anteroposterior and medial-lateral translations or the rotations in any plane. To assess soft tissue artifact (STA), error was calculated via root mean square error (RMSE) and linear fit method (LFM) between DSX and OMC data. DSX and OMC average R^2 value between sockets had strong agreement in the sagittal plane (0.928) in contrast to the medial-lateral and axial planes (0.431; 0.218) which had large disagreements in wave form similarity.

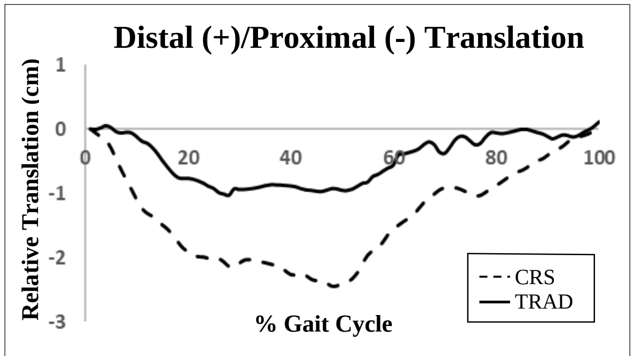

Figure 1: Proximal-Distal translation for each socket type. The traditional socket had significantly less axial translation compared to the CRS socket.

DISCUSSION

The stabilization of the residual limb is essential to optimize prosthetic function and comfort. Ensuring a proper socket fit is partially predicated on the analytical and experimental tools that aid in modeling bone pose within a prosthetic socket. DSX is currently the only available technology that can achieve sub-millimeter bone position and orientation (pose) estimation accuracy during a wide variety of functional movements. This investigation demonstrated that the CRS socket design had significantly increased axial translation compared to the TRAD socket for individuals with TFA during treadmill walking at self-selected speed. LFM and RMSE models further reinforced OMCs limited capabilities of tracking residual limb movement during gait. The ability to accurately assess the dynamic interaction between the residual limb and socket is necessary to develop effective, evidence-based prosthetic solutions to reduce secondary physical comorbidities and degenerative changes that result from complications of poor prosthetic load transmission.

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DISCLOSURE STATEMENT

The contents of this publication do not state or reflect the views of the United Stated government and shall not be used for advertising or product endorsement. There was no COI.

Squatting Kinematics in Patients with Unilateral and Bilateral Acetabular Hip Dysplasia

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INTRODUCTION

Acetabular Hip Dysplasia (AHD) refers to an abnormal skeletal development of the hip with a range of abnormalities including deficient shape, size and/or orientation. The irregular acetabulum leads to poor femoral head coverage, an increase in mechanical stress, and degenerative changes of the joint [1].

A traditional gait assessment may not be sensitive enough to exemplify clinical differences in patients with AHD. Additional functional tasks such as a squat become necessary to analyze due to the required hip mobility of the movement. Poor mobility may cause various compensations, such as an increased trunk lean [2]. A previous study analyzed patient reported outcomes on patients clinically diagnosed with AHD; two-thirds of the patients reported having pain while sitting in a chair [3]. The squat is closely linked to activities of daily living, such as descent and ascent from in a chair, bending over to pick up children, lifting packages off the floor, etc. The purpose of this study was to compare three different squat methods in patients with unilateral and bilateral symptomatic AHD and assess alterations in hip biomechanics.

CLINICAL SIGNIFICANCE

Understanding the squat biomechanics in patients with hip disorders can assist in developing appropriate rehabilitation and treatment plans to improve function.

METHODS

Data collected through an IRB-approved hip preservation surgical registry was reviewed to identify adolescent and young adult patients clinically diagnosed with AHD with no neurologic or syndromic abnormalities (data pulled from 2012-2019). The current analysis was conducted on thirty-three (N=33) pre-operative patients (31 females, aged 17±4yrs., height 162±9cm, mass 61±16kg).

For all squats, patients were instructed to stand with their feet shoulder-width apart, arms extended straight out in front with hands pronated. The series of squats consisted of the hold squat, standard squat and target squat. During the hold squat the patients were instructed to squat down to their lowest comfortable point, hold this position for a count of three, and then return to the upright standing position; standard squat – patients were instructed to squat down to their lowest comfortable point then immediately return to their original upright standing position; target squat – instructions were similar to the standard squat with the addition of the 15.5 cm bench placed behind the heels of the patient [4]. Events for all three types of squats (Hold, Standard, Target) were determined using a custom MATLAB code [5]. A representative squat trial was chosen for each squat based off of maximal squat depth.

RESULTS

There was no significant difference between unilateral (n=15) and bilateral (n=18) patients across all variables ($p>0.05$). Ultimately, patients accomplished their deepest squat depth during the target squat and there was a significant difference in squat depth between the standard and hold squat (7±7%, $p<0.001$), target and standard squat (9±8%, $p<0.001$), and target and hold squat (16±9%, $p<0.001$). There was a significant difference in trunk flexion between the standard and the hold squat (2±6°, $p=0.033$), target and standard squat (7±8°, $p<0.001$), and target and hold squat (9±9°, $p<0.001$) with the most trunk flexion in the target squat (37±13°). Furthermore, there was greater hip flexion in the target squat (102±18°) compared to the standard and hold squat with a significant difference between the standard and hold squat (5±6°, $p<0.001$), target and standard squat (5±7°, $p<0.001$), and target and hold squat (11±9°, $p<0.001$). There was a significant difference in knee flexion between the standard and hold squat (12±11°, $p<0.001$), target and standard squat (11±12°, $p<0.001$), and target and hold squat (23±13°, $p<0.001$) with the most knee flexion in the target squat.

DISCUSSION

Based on the preliminary results of this study, it is clear that there are different kinematic strategies for each squat type in patients with AHD (Table 1). Participants achieved considerably greater squat depth in the target squat. Furthermore, trunk flexion, hip flexion, and knee flexion all increased in the target squat compared to the standard and hold squat. While hip flexion increased from the standard squat to the target squat, the greatest increase was in knee flexion. These results may imply both the target and hold squat may be most advantageous during rehabilitation in order to achieve maximal squat depth as well as gain the stability necessary for activities of daily living. Future work should focus on investigating the relationship of ankle dorsiflexion, step width, and foot progression angle.

Surgical Limb	**Group Comparison** mean (SD)			**Hold vs Standard**	**Target vs Hold**	**Target vs Standard**
	Hold	**Standard**	**Target**	***p*-value**	***p*-value**	***p*-value**
Max Trunk Flexion (°)	28.1 (12.8)	30.5 (12.1)	37.3 (13.2)	**0.033**	**<0.001**	**<0.001**
Pelvic Obliquity (°)	3.0 (3.8)	2.9 (4.1)	3.3 (4.3)	0.693	0.362	0.102
Max Hip Flexion (°)	91.5 (17.3)	96.7 (16.6)	102.0 (17.8)	**<0.001**	**<0.001**	**<0.001**
Max Hip Abduction (°)	17.4 (7.1)	17.5 (7.3)	17.1 (8.4)	0.911	0.716	0.621
Max Knee Flexion (°)	95.2 (22.7)	107.1 (22.5)	118.2 (20.8)	**<0.001**	**<0.001**	**0.003**
Max Squat Depth (%)	39.0 (13.8)	46.5 (13.8)	55.2 (13.8)	**<0.001**	**<0.001**	**<0.001**

Table 1: Sagittal and coronal plane kinematics of the affected limb. Mean (SD) and statistical comparisons of the three squat types.

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DISCLOSURES

The authors have no disclosures to report

Consideration of mucopolysaccharidosis as a differential diagnosis for patients with skeletal dysplasias of the hips seen in the motion lab: a case study

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Introduction
Mucupolysaccharidoses (MPS) are a group of rare, inherited autosomal recessive multisystem disorders with varied presentations caused by enzyme deficiency.[1,2] This enzyme deficiency causes a buildup of glycosaminoglycans, or large sugar molecules, in the lysosomes of bones, joints and organs.[2,3] Patients with nonclassical phenotypes may have skeletal issues that resemble other conditions such as spondyloepiphyseal dysplasia (SED), multiple epiphyseal dysplasia (MED), undiagnosed skeletal dysplasia, pseudoachondroplasia and Legg Calve Perthes Disease (LCPD). Due to lack of knowledge of this disorder, diagnosis of MPS may take months to years to be established and a misdiagnosis is often given. Treatment for MPS often includes enzyme replacement therapy and possibly a bone marrow transplant to help manage the symptoms.

Clinical Significance
The current case study describes the gait deformities, muscle weakness, activity and participation impairments, surgical interventions, therapy and journey to an MPS diagnosis for a teenage patient who was seen by many specialists over a 7 year period thought to have LCPD, SED, and/or MED. We hope to educate others on MPS as a possible differential diagnosis for skeletal disorders so that their patients might be able to receive timelier genetic testing and enzymatic replacement and management.

Methods
This was an otherwise healthy 11yo African male with bilateral hip pain, hip dysplasia with collapse of the epiphyses and difficulty walking with an antalgic gait for the past three years when seen initially at our institution. He had seen several other specialists in orthopedics, neurology and physical therapy (PT) and had undergone radiology, blood work, MRI, EMG and nerve conduction studies. Prior to our initial visit he was thought to have LCPD and underwent PT without success. He was worked up for thrombophilia and other disorders and was then thought to have MED or SED. The patient was able to walk on his own with a significant Trendelenberg gait and flexed hips for short community distances. He often crawled at home. Significant weakness of the hip abductors and extensors was noted and he developed a thoracic scoliosis. At age 11 he underwent a left hip Dega acetabuloplasty, proximal femoral varus osteotomy and adductor longus tentotomy. At age 14 he underwent R proximal femur varus producing osteotomy and left hip hardware removal. Despite extensive PT, he was never able to gain back normal strength of the right hip and developed some right knee pain but he remained active walking at school and participating on the basketball team. Continued activity difficulties included poor posture in standing, fatigue with running, inefficient gait mechanics, difficulty squatting/jumping/stair negotiation and rising from the floor. He continued to be followed by the orthopedic and neurology service and had several gait videotapes and a full gait study at the age of 13 just prior to his right hip surgery. At the age of 15, the diagnosis continued to be questioned and he was referred for lab work to evaluate for MPS enzyme deficiency due to his symptoms of hip dysplasia and scoliosis. Testing confirmed a diagnosis of MPS IVA or Morquio syndrome and he was referred to a MPS genetics clinic for possible enzyme replacement therapy.

Results
Prior to his left hip reconstruction, video analysis demonstrated marked trendelenberg; increased lordosis; shortened left stride length; decreased left stance time; and persistent left hip flexion with pelvic retraction. Instrumented 3D gait analysis was performed 2 years following left hip reconstruction. Kinematic results included increased trunk motion in all planes; pelvic asymmetry with elevation and

protraction of the left hemipelvis; diminished right hip extension; increased left hip external rotation and increased bilateral hip abduction (Figure 1). Kinetic data showed decreased bilateral hip abduction moments and decreased hip flexor moments in terminal stance. Left hip power generation was increased at loading response and hip power generation was decreased at terminal stance right greater than left (Figure 2). The gait data objectively confirmed the observational gait deviations of suspected hip weakness, pelvic obliquity, pelvic rotation, and trunk asymmetries.

Discussion

The clinical signs of MPS or Morquio syndrome are so varied that this may be missed as a differential diagnosis.[5] Patients with skeletal issues such as hip dysplasia, hip pain and stiffness with associated weakness and gait dysfunction seen in motion lab centers are often diagnosed with some type of hip dysplasia when in fact they may have a lysosomal storage disorder. We hope this case study from the motion lab helps to bring increased awareness of early genetic testing of MPS for patients seen in gait labs and orthopedic clinics who have non-classical musculoskeletal presentations. Earlier testing and treatment of this enzyme deficiency could lead to improved patient outcomes, quality of life and management of the disease manifestations.

Figure 1: Pelvic and hip kinematics

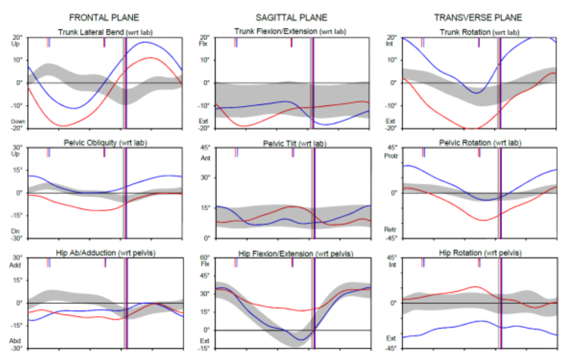
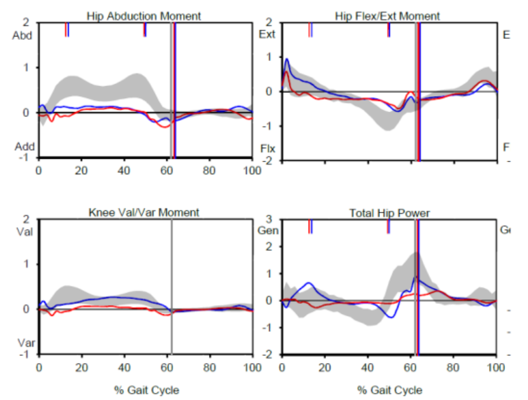


Figure 2: Pelvic and hip kinetics



Key to graphs: red = right, blue = left

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DEVELOPMENT AND TESTING OF 3D-PRINTED LOWER LIMB PROSTHETIC SOCKETS

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INTRODUCTION

Fabrication methods for lower limb prosthetic sockets has remained nearly static for decades - despite it being time-consuming, messy, somewhat expensive, and labor intensive. Advances in 3D printing technology present the possibility to reduce fabrication time, labor and cost, make the process more uniform and objective, and provide a perfect digital copy for any subsequent fabrications. Some preliminary work has begun to be published about this process and the resulting product but many methodological and performance questions remain. Computer modeling (CAD) design choices, material selection and 3D printing parameters can significantly influence comfort, safety, fabrication time and cost. The present work attempts to shed light on the impact of these choices. In this paper, we describe initial findings on the effect of material and 3D printing parameter choices on print speed, post-fabrication modification (i.e. workability) and strength of 3D printed sockets.

CLINICAL SIGNIFICANCE

Though a few commercial services exist for 3D printed socket fabrication, the basic socket design methodology, choices associated with materials and final socket fit and structural properties have not been scientifically, if at all, well described. Clarifying the role and impact of these measures is essential to ensure a safe and appropriate socket can be delivered to patients, to develop consensus on best practices and to quantify the benefits of this new approach.

METHODS

To gauge the suitability of print materials on workability (i.e. post-fabrication modifications), sample of 8 common 3D printing materials were cut, heated, flared and buffed by a CPO – exactly as is done on sockets during fitting and delivery. A qualitative rating was then given by the prosthetist to each material. We tested PLA (Polylactic Acid), CPE (Co-Polyester), PVA (Polyvinyl alcohol), PP (Polypropylene), ABS (Acrylonitrile Butadiene Styrene), PETG (Glycol-modified Polyethylene Terephthalate), PC-MAX (Polycarbonate) and HIPS (High Impact Polystyrene). Plates (2in x 2in x 0.25in) of each material were printed at 30% infill. To gauge the effect of printing parameters on socket fabrication time, print time estimates by the slicing software were recorded as material, infill percentage, printing pattern, printing orientation, and layer height were varied. Initially, sockets were actually printed while the independent parameters varied. It was found that the slicer time estimates were accurate to within 5% of actual print times, which was deemed sufficient for this work. All subsequent time data were obtained print simulations but not actually printing the sockets in those configurations. Jigs were designed and machined to allow proper loading to be applied by a testing machine (Tinius Olsen, H10KS,

Fig. 1 Strength testing setup.

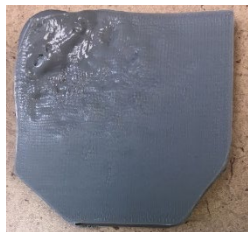
Horsham, PA USA). Per general guidelines adapted from ISO10328 (Conditions I&II, static load test), sockets were to be tested at various angles and loads to represent those that may occur during walking. The experimental setup is shown (Fig.1). To evenly distribute the load, the socket was filled with a polymeric sand before loading was applied. Since variability in the strength of conventional (and a few 3DP) sockets is emerging [1-4], we will follow the core testing techniques described to date to facilitate comparison to those results. Sockets will be tested to proof and ultimate failure levels. This testing is currently ongoing.

RESULTS

The material workability evaluations (Table 1) and photos (Fig. 2) help clarify these results. Testing for the other four materials is underway. The effect of print settings on socket fabrication times is reported in Table 2.

Table 1. Material Workability Ratings

Material	Workability	Heat Tolerance
PLA	Poor	Poor
CPE	Great	Good
PVA	Poor	Poor
PP	Good	Great



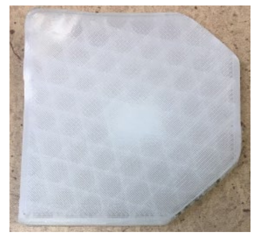


Fig 2. Left) PLA – showing warping during heating; Right) PP – showing excellent heat tolerance and good workability

DISCUSSION

We believe this is the first report on the effect of print settings and materials on the properties of 3DP sockets. It serves am-putees to have well described specifications that are evaluated and debated by care providers, engineers and patients – to reach consensus on best practices. We have benchmarked the effect of numerous printing parameters (layer height, material, infill percentage, printing pattern, printing orientation) – and determined that layer height has one of the most significant impacts on print speed. As it stands this has potential to reduce socket delivery times considerably. Further improvements may allow sockets to be ready the same day – significantly reducing travel burden on the patient. Our testing has shown that one of the most common 3DP materials, PLA, is not suitable for sockets but other fairly common materials (PP, CPE) may well be – pending strength testing. Other materials are still being evaluated. Strength testing is in early stages but shows our general method for design and fabrication produces high quality strong sockets (static testing loads>1000lbs). Subjective fit testing of socket to residual limb has been very positive.

Table 2. How Fabrication (3D Print) Times Vary with Print Settings

Material	Infill %	Pattern	Layer Height (mm)	Orienta-tion	Time (hr:min)
PLA	20	Triangle	0.3	Vertical	10:05
PLA	40	Triangle	0.3	Vertical	13:33
PLA	20	Triangle	0.2	Vertical	17:13
PLA	20	Grid	0.3	Vertical	08:39
PETG	20	Grid	0.2	Vertical	12:42
PETG	20	Grid	0.2	45 degs.	17:52

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Long-term Evaluation of Kinetics and SF-36 Scores after Intramedullary Nailing of Tibial Shaft Fractures

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INTRODUCTION

Tibia fractures occur at an incidence of approximately 16.9/100,000 per year [1]. Tibial diaphyseal fractures are commonly treated with intramedullary nailing. Functional outcome studies have shown that 47-86% of patients continue to experience anterior knee pain even after the bone is fully healed [2]. Long-term function of patients that underwent surgical fixation of tibial shaft fractures has not been explored using 3D gait analysis correlated with outcomes assessments. The goal of this study was to analyze the gait kinetics of persons at least 2 years after an operatively managed tibial shaft fracture compared to control data. It was hypothesized that there would be no alterations in the gait kinetics between the study and control data.

CLINICAL SIGNIFICANCE

Characterization of kinetics during ambulation following intramedullary nailing for tibial shaft fractures along with qualitative Short Form 36 (SF-36) responses provide valuable insight to better understand long-term postoperative outcomes, including anterior knee pain.

METHODS

Sixteen participants (10M, 6F, 39.81 ± 14 years; 78.71 ± 13.84 kg) at least two years (5.9 ± 2.5 years) post-operative from a tibial diaphyseal fracture treated with intramedullary nailing consented to participate in our IRB-approved protocol. Following written informed consent, each subject completed the SF-36. Gait analysis was performed using 12 Vicon cameras and Nexus software system. Each participant was fitted with 17 reflective markers placed on key anatomical landmarks. Each participant walked across a 30-foot walkway at a self-selected pace for 10 good trials. Temporal-spatial gait parameters, categorical SF-36 scores, and kinetics were analyzed. Results from this population were compared to data from a control population (Controls) of 26 young adults with no lower extremity pathologies (13 M, 13 F; 30.4 +/- 4.5 years). Welch's two-tailed t-tests were used to assess statistical significance at an alpha level of 0.05.

RESULTS

SF-36 scores from the study population revealed poor satisfaction and performance compared to the controls except in the "Role limitations due to physical health" category (Table 1). The study population had a statistically significant decrease in walking speed compared to control data (1.04 m/sec vs. 1.21 m/s; $p<0.01$). Kinetic plots for significant differences between the two data sets are shown (Fig. 1). The study population demonstrated significant decreases in ankle power ($p<0.05$) from 42-68% of the gait cycle. There was a statistically significant ($p<0.04$) decrease in the ankle flexion/extension moment demand compared to control data during stance phase following initial contact. The study population also exhibited a statistically significant difference ($p<0.05$) in the knee flexion/extension moment for the majority of stance phase. Large effect size (d>0.8) was calculated for knee and ankle flexion/extension moments of the study population during the stance phase of the gait cycle (4-58% and 2-62%, respectively).

Table 1. SF-36 results of the study and healthy normal population and the Cohen's d effect size between populations. $^{+}$represents a moderate effect and * represents a large effect.

Category	Fracture	Control	Cohen's d Effect Size
Physical functioning	90.67	97.64	$^{+}$0.5
Role limitations due to physical health	98.33	97.92	0.2
Role limitations due to emotional problems	97.78	97.92	0.1
Energy/fatigue	71.67	98.06	0.2
Emotional well-being	86.2	98.33	0.3
Social functioning	95.83	98.06	0.4
Pain	81.67	97.92	*1.4
General Health	77.33	97.78	$^{+}$0.6

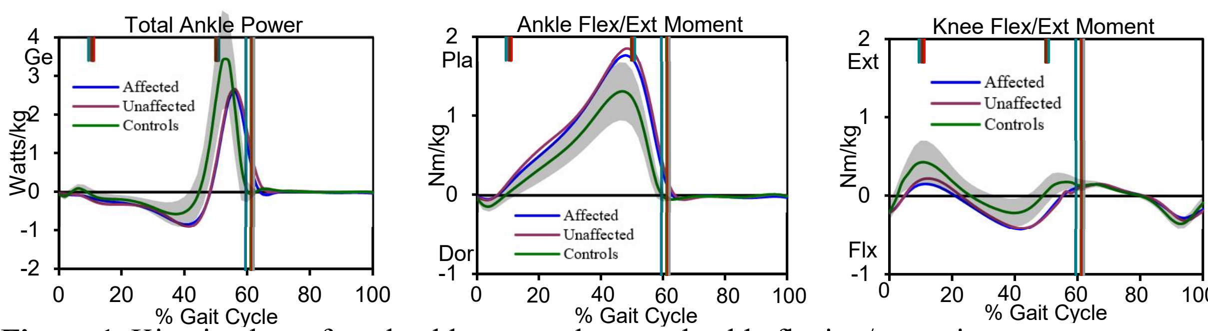

Figure 1. Kinetic plots of total ankle power, knee and ankle flexion/extension moments.

DISCUSSION

Ankle push off power and knee and ankle flexion/extension moments were significantly altered in the fracture population. The study population demonstrated a flexion moment demand shift in the knee throughout stance when clearing the ground. Ankle moments were also affected and showed an increased plantar flexion moment demand. According to SF-36 results, the study population rated their energy/fatigue, pain, and general health to be lower than the healthy normal population. Therefore, it is likely that the decreased walking speed and the above-mentioned deviations of gait may be a result of decreased energy and increased pain as reported by the study population. Limitations of this study include small sample size.

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DISCLOSURE STATEMENT

G. Schmeling is on the editorial board of The American Journal of Orthopaedics, The Journal of Orthopaedic Trauma, and The Journal of Surgical Education. All other authors have no conflicts of interest to disclose.

Alterations of Plantar Pressure and Migration of Pressure During Gait With a Prefabricated Pneumatic Walking Brace

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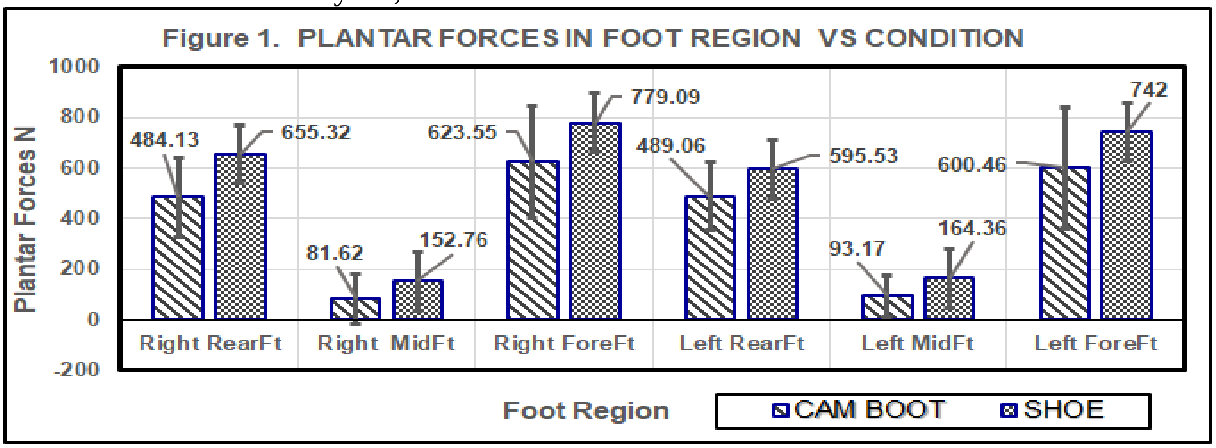
Introduction: Lower extremity injuries can have major and long lasting effects on gait. A total of 280,933 foot/ankle fractures or dislocations were identified in the US from 2007 – 2011. [1] Changes in gait patterns can lead to forces being applied on the body in a direction that do not normally occur. Alterations in weight bearing during perturbed gait may expose the regions of the foot to a higher risk of injury. Gulgin H. et al.[2] conducted a kinematic study in which females and males wore a boot on the right foot and a tennis shoe on the left. The subjects' velocities decreased with the walking boot, and increased all 3D planar motion in the pelvis, hip and knee joints. They concluded the walking boot created asymmetries in both lower extremity joint angles, which may result in secondary pain in more proximal regions of the body. Baumhauer et al.[3] examined the plantar pressure applied on the foot when comparing a standardized shoe, prefabricated pneumatic walking brace, and total contact cast. They reported that the prefabricated pneumatic walking brace decreased peak plantar foot pressures to an equal or greater degree than the total contact cast in all tested locations of the forefoot, midfoot, and hindfoot. The current study compared the foot pressures of the plantar aspect of the foot when an individual is walking in an Aircast prefabricated pneumatic walking brace versus walking in a New Balance shoe. Tekscan F-scan foot sensors were used to record foot pressures in the rearfoot, midfoot, and forefoot regions to examine alterations of plantar forces and the migration of foot pressure when walking shod and with a brace.

Clinical Significance: Examine the effects of the Aircast cam boot on the foot plantar foot regions measured by a Tekscan F-scan sensor and compare it to a shod condition during normal walking gait to examine if abnormal force applications occur on the foot while wearing a boot which may lead to further injury or perturbed gait.

Methods: Seven female and eight male subjects between the ages of 18 and 24 years participated in the study. Participants completed a medical history survey for the following exclusionary criteria: previous lower extremity or back injury within the last 6 months. This study was approved by Indiana State University IRB. Subjects were provided a new pair of New Balance shoes for the shod condition which was equipped with a calibrated Tekscan FScan foot pressure sensor. Also, prior to the subjects' gait trials with the Aircast boot equipped with the F-scan sensor, they performed 10 minutes of boot walking habituation. Then the subjects walked on a flat 9m walkway. Plantar data for the right and left strides for the shod and cam boot conditions were collected for 2 walking trials after the 3m marker. Infrared timing gates were placed at the 3m and 6m marker for the measurement of gait speed. To ensure a consistent gait speed, the subjects' gait speed had to fall within the range of 1.3 to 1.5 $m*s^{-1}$ in order for the trial to be accepted for analysis. Subjects continued to perform walking trials until two acceptable trials were completed. The exact same protocol was then repeated until two acceptable trials with a walking boot on the left foot and right foot were collected at 100 Hz using the Tekscan F-scan software.

Results: A 5-way ANOVA (Boot/Shoe x Ft region x Foot x Step x Trial) was performed to assess the significance of the pneumatic walking boot on plantar pressures across the various regions of the foot. When examining the boot versus shoe condition, a significant p-value of .000 was found to exist. The analysis found that the variances in plantar forces for foot regions (rearfoot, midfoot, forefoot) produced a significant p-value of .000 for the different foot regions which is shown in Figure 1. In order to determine that our data was reliable and repeatable, the statistical analysis compared the variances between right and left leg data, trial 1 and 2 data, and step 1 and 2 data, all conditions resulted in non-significant p-values. These results indicated that the plantar pressures were similar between the right and left foot contacts, between the first and second stride analyzed, and between the 2 trials selected for each condition.



Discussion: Significant changes in the distribution between the normal walking shod condition and the boot condition were found. The shod condition resulted in the overall plantar foot surface having 45% more force applied than when walking with the cam boot. Significant differences existed between plantar pressures on the different regions of the foot between wearing shoes, and wearing a shoe on one foot and an Aircast prefabricated pneumatic walking boot. A two-way interaction between the condition and foot region data was found to trend near significance. The shod rearfoot received 28.6% more force, shod midfoot 81.8% increased force, and shod forefoot had 24.3% more force applied to this foot region. The reduction of foot forces when walking with a boot would be beneficial for recovery from an injury. When examining the timing of the transition of forces across the different regions of the foot, differences were noted between the boot and shoe conditions. Future kinetic and kinematic studies should be conducted investigating the influence of the 90 degree ankle axis of rotation in the cam prefabricated pneumatic walking boot as compared to the 16 degree oblique angle ankle axis found naturally in the foot during the gait pattern.

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Comparison of Ankle Kinematics between a Multi-segment Foot Model and a Single-segment Lower Extremity Model in the Context of Total Ankle Arthroplasty

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INTRODUCTION

Total ankle arthroplasty (TAA) is becoming the standard of care for end stage ankle arthritis because it reduces pain while preserving ankle motion [1-3]. A key to improving patient outcomes is the ability of TAA to normalize ankle dorsiflexion during early-stance phase and plantar flexion during late-stance phase [1]. Typical gait analysis models the foot as a rigid segment [2,3]; however, this approach may lead to inaccuracies in ankle joint kinematics [4]. The Milwaukee Foot Model (MFM) is a multi-segment foot model that represents the foot and ankle complex as four segments (hallux, forefoot, hindfoot, and tibia) to provide a more accurate depiction of foot motion during gait [5]. The purpose of this study is to determine if the sagittal gait kinematics differ between the MFM and a single segment model, specifically in the context of ankle arthroplasty.

CLINICAL SIGNIFICANCE

The choice of biomechanical model should be considered when assessing gait kinematics for patients who have undergone total ankle arthroplasty.

METHODS

Six participants who had received the STAR (Scandinavian Total Ankle Replacement, Stryker, Kalamazoo, MI) implant were evaluated an average of 31 months (range 11-68 months) post-operatively per an IRB-approved protocol. After consenting to the study, participants were fitted with reflective markers for the MFM and Vicon's lower extremity Plug-in Gait (PiG) model (Vicon Motion Systems, Ltd.; Oxford, UK). Participants walked barefoot along a 30-foot marked path at a comfortable, self-selected pace, while twelve infrared cameras collected data for a minimum of ten trails. Ankle kinematic data from the two model outputs were compared. Welch's t-tests were used for statistical significance calculations. Cohen's *d* effect size was used to determine minimal clinically important differences.

RESULTS

Statistically (maximum p-value <0.02) and clinically (minimum Cohen's d >2.0) significant differences were seen throughout the entire gait cycle when comparing sagittal motion of the PiG ankle joint to the MFM hindfoot segment. On average, the MFM hindfoot was 15.7° more dorsiflexed than the PiG ankle, with a maximum difference of 19.1° at the end of swing phase and a minimum of 12.4° at the end of stance phase (Fig. 1). The PiG ankle averaged 4.0° of dorsiflexion at the ankle (range -2.5° [plantar flexion] to 12.8°) and the MFM hindfoot averaged 19.7° of dorsiflexion (range 12.6° to 25.1°), while the forefoot segment of the MFM averaged 37.1° of plantar flexion (range of 33.1° to 41.7°).

DISCUSSION

The results of this study show that the PiG model's sagittal ankle angle is statistically and clinically different from the MFM's hindfoot sagittal angle throughout the gait cycle. Because the PiG models the foot using markers on both the hindfoot (calcaneus) and forefoot (head of 2nd metatarsal), it combines the motion of these two segments. The MFM's multi-segmental representation allows for the motion of each segment to be calculated separately and its radiographic indexing to the bony foot anatomy further improves accuracy. A summation of the forefoot and hindfoot sagittal motions yields a combined angle comparable to the PiG model; however, this total angle may misrepresent the true motion at the tibiotalar joint (Fig. 1).

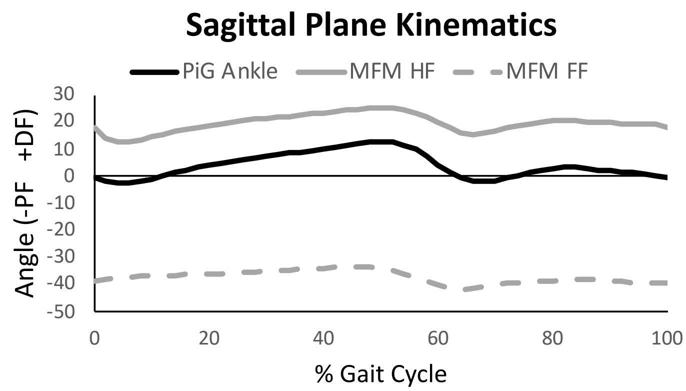

Figure 1. Sagittal plane kinematics over the gait cycle for the PiG ankle (foot with respect to tibia), MFM hindfoot (HF) with respect to tibia, and MFM forefoot (FF) with respect to hindfoot segments.

Most gait analyses of post-operative TAA patients use a single-segment model to determine sagittal ankle motion [1-3]. Singer et al. reported a range of 6.2° of plantar flexion to 11.9° dorsiflexion and Queen et al. reported a range of 4.2° of plantar flexion and 7.4° of dorsiflexion, both similar to the results seen with the PiG model in our study (2.5° plantar flexion to 12.8° dorsiflexion) [1,3]. These differences compared to the MFM are to be expected because single-segment modeling includes the distal part of the foot when calculating sagittal motion of the ankle and does not distinguish between forefoot and hindfoot motion. Therefore, it is important to recognize the biomechanical model used when comparing kinematic data across studies. Future studies should consider using a multi-segment foot model and radiographic indexing to more accurately represent sagittal motion of the tibiotalar joint. In conclusion, modeling of the foot should be carefully considered when calculating and comparing gait kinematics for patients who have undergone TAA.

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DISCLOSURE STATEMENT

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No other authors have conflicts of interest to disclose.

MULTI-SEGMENT FOOT COORDINATION IN PEDIATRIC PATIENTS WITH PLANOVALGUS FOOT DEFORMITY

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INTRODUCTION

The development of multi-segment foot models presents new opportunities to better understand the structure and function of the foot in patients with neuromuscular impairments. Vector coding techniques based on relative motion plots can be used to quantify the coupling angle between two segments or joints. Vector coding has been used to describe the coordination of foot segments/joints in healthy adult participants[1-3] and in one case study of a patient with club foot.[4] The purpose of this pilot study was to compare foot coordination patterns during walking in typically developing children (TD) and children with cerebral palsy and planovalgus foot deformity (PV).

CLINICAL SIGNIFICANCE

An understanding of the coordination of foot segments during gait may provide information useful for the treatment planning of pediatric planovalgus foot deformities.

METHODS

This retrospective analysis was based on a convenience sample of patients who 1) had undergone an instrumented 3D gait analysis in our laboratory 2) presented with a clinically identified planovalgus foot, 3) had a diagnosis of cerebral palsy, and 4) walked with a crouch gait pattern. Kinematic data collected on typically developing children as part of our normative database were also utilized. All patients (PV) and typically developing participants (TD) had completed a gait analysis without the use of ambulatory aids following a standardized protocol that included use of the multi-segment IOR foot model.[5]

A modified vector coding approach described by Needham[3, 6] was used to classify the coupling angles as in-phase proximal dominancy (IPP), in-phase distal dominancy (IPD), anti-phase distal dominancy (APD) and anti-phase proximal dominancy (APP). The coupling angle for five pairs of joint rotations were calculated: 1) calcaneus-midfoot inversion (CM Inv), 2) calcaneus-midfoot dorsiflexion (CM Dorsi), 3) midfoot-metatarsus inversion (MM Inv), 4) metatarsus-hallux dorsiflexion (MH Dorsi), and 5) calcaneus-midfoot abduction (CM Abd) each relative to shank-calcaneus inversion (SC Inv). The coupling angles were calculated over the stance phase of the gait cycle which was divided into 3 sub-phases: early stance (0-33%), mid stance (34-66%) and late stance (67-100%).[1, 2, 7] For this preliminary analysis, the average coupling angle during each period of stance was calculated for each participant using a circular average approach.[7, 8] Then a Watson-Williams test was used to compare the average coupling angle across groups for each of the three stance phases for each joint pair with significance set at $p < 0.05$.[7, 8]

RESULTS

The data from 9 TD participants (6 male and 3 female) ages 9 ± 3 years old and 6 PV participants (2 male and 4 female) ages 12 ± 4 years old are presented. There were significant differences in average coupling angles for the MM Inv vs. SC Inv and CM Abd vs SC Inv coordination pairs, Table 1. The mean coupling angles across stance for each group for these joint pairs are shown in Figure 1. No other statistically significant differences were found.

Table 1: Mean Coupling Angles (mean± SD).

	Early Stance			**Mid-Stance**			**Late-Stance**		
	PV	TD	P	PV	TD	P	PV	TD	P
MM Inv vs SC Inv (deg)	**106 ± 56**	**277± 64**	**0.0001**	**94 ± 48**	**37 ± 53**	**0.0325**	15 ± 44	355 ± 26	0.1927
CM Abd vs SC Inv (deg)	93 ± 36	79± 65	0.5532	**90 ± 69**	**342 ± 60**	**0.0193**	**340 ± 47**	**298 ± 13**	**0.0049**

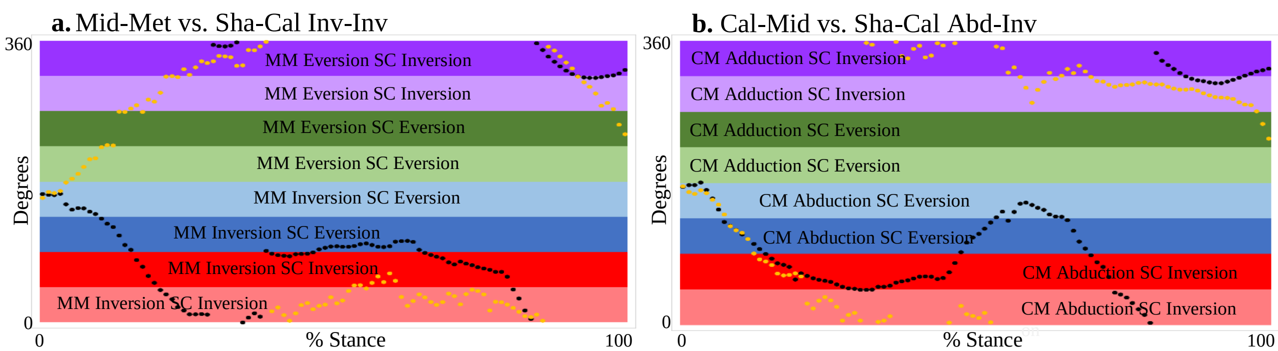

Figure 1. a) Mean coupling angle during stance for MM Inv vs. SC Inv; b) Mean coupling angle for CM Abd vs SC Inv. ■APP ■APD ■IPD ■IPP ■APP ■APD ■IPD ■IPP •PV •TD

DISCUSSION

The vector-coding analysis used in this study revealed different foot segment coordination patterns between the TD and PV groups. The vector-coding approach proposed by Needham[3, 6] allowed for a clinically meaningful interpretation of the results. For example, in early stance the coordination patterns for the MM Inv and SC Inv are consistent with midfoot collapse during weight acceptance often observed in those with planovalgus foot deformity. These preliminary results support the potential of multi-segment foot coordination in the analysis of foot function during gait in patient populations.

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DISCLOSURE STATEMENT

Authors have no conflicts of interest to disclose.

CARBON FIBER FOOTPLATE WEAR IMPROVES GAIT KINEMATICS OF CHILDREN WHO IDIOPATHICALLY TOE WALK

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INTRODUCTION: The development of a functional gait pattern is essential to participating in activities of daily living, facilitates positive social interactions, and is influential in measuring and improving quality of life[1]. Gait assessments can help therapists understand typically developing gait, identify impairments, and predict motor outcomes in children[2]. Studies gathering normative data for typically walking children have shown children take on an adult-like gait pattern between the ages of 7 and 11 years[3].

Idiopathic toe walking (ITW) is a phenomenon of unknown etiology[4] that can affect children as young as 18 months, and in severe cases when left untreated, can persist into adolescence[5]. 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[6].

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 and provide a sensation of deep pressure[7] that may influence their sensory seeking needs[8]. The current study highlights the long-term efficacy of CFO intervention in altering the gait kinematics of children who walk on their toes at an age that is on the threshold of transitioning to an adult-like gait pattern.

CLINICAL SIGNIFICANCE: ITW is a difficult condition to treat in an estimated 7-24% of children[9]. CFOs are an affordable and convenient intervention with potentially positive impacts on gait in children with ITW. Benefits to clinicians include providing conservative and risk-free treatment techniques before discussing more invasive interventions such as Botox injections or Achilles lengthening surgery.

METHODS: 10 children ages 3 to 8-years-old (6.3 ± 1.4 years) whose parents reported toe walking and no neurological diagnosis, surgical history or current orthoses participated in this study. Data were collected at baseline and repeated at a 6-week follow-up visit. Active and passive ankle dorsiflexion, hamstring and quadricep flexibility were measured using goniometry. Participants walked at a normal pace on an instrumented Zeno walkway system for 5 laps with no CFOs and 5 laps with CFOs inserted beneath bilateral insoles. Average foot contact area, stance %, and velocity were measured. All walking trials were video recorded in the sagittal plane. Utilizing Dartfish video analysis software, joint angles of hip extension were measured in terminal stance, and knee flexion, and plantar flexion were measured at initial contact. Participants were asked to wear the CFOs for 6 weeks during typical activities. Static flexibility data were analyzed using paired t-tests and two-way repeated measures ANOVA were performed to compare condition of CFO use before and after intervention.

RESULTS: When comparing conditions, using CFOs resulted in significant increase in foot contact area (P=0.017; Table 1), hip extension (P=0.027; Table 2), and decrease in plantar flexion (P=0.004; Table 2). After 6 weeks, knee flexion at terminal stance significantly increased with the CFOs donned (P = 0.047; Table 2). Static measures of active range of motion (ROM) ankle dorsiflexion (Left P=0.030, Right P=0.001) and quadriceps flexibility (Left P=0.003, Right P=0.007) significantly increased after using CFOs for 6 weeks.

Table 1: Mean ± SD Temporospatial Measurements with Zeno Walkway

	No CFO Baseline	CFO Baseline	No CFO at 6 wks.	CFO at 6 wks.
Foot contact area (cm^2)	133.24 ± 12.67	148.39 ± 8.83	145.08 ± 8.86	149.31 ± 9.20
Stance %	59.07 ± 0.25	59.61 ± 0.54	58.68 ± 0.31	59.99 ± 1.20
Velocity (cm/sec)	129.41 ± 7.48	131.09 ± 6.41	129.07 ± 3.25	121.98 ± 6.60

Table 2: Mean ± SD Joint Angles during Gait Measured with Dartfish App

	No CFO Baseline	CFO Baseline	No CFO at 6 wks.	CFO at 6 wks.
Hip Ext (°)	4.12 ± 0.64	4.80 ± 0.78	3.45 ± 1.10	4.16 ± 0.98
Plantar Flex (°)	51.73 ± 2.83	41.08 ± 2.28	45.31 ± 26.05	44.11 ± 2.72
Knee Flex (°)	52.64 ± 1.65	55.52 ± 2.86	56.33 ± 0.94	57.05 ± 1.10

DISCUSSION: Children who walk on their toes showed improvement in gait kinematics and joint ROM in static and dynamic assessments after using CFOs for 6 weeks. These changes may lead to decreased pain and increased function in children with ITW, and assist in progression towards a typical adult gait pattern upon maturation.

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CLINICAL EFFICACY OF INSTRUMENTED GAIT ANALYSIS: SYSTEMATIC REVIEW 2019 UPDATE

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INTRODUCTION

In 2011, we published a systematic review evaluating the literature related to the clinical efficacy of 3D instrumented gait analysis (3DGA) [1]. This topic is still important since debate persists regarding the use of 3DGA for clinical patients and with the increasing emphasis on evidence-based medicine. The current study is an update to our previous review.

CLINICAL SIGNIFICANCE

Evidence of efficacy is needed to support the utilization and reimbursement of instrumented motion analysis in clinical care.

METHODS

A literature review was conducted to identify English language articles related to human gait analysis published from September 2009 to October 2019. Papers had to investigate human walking using 3D kinematics, kinetics, ground reaction force, plantar pressure, and/or electromyography (EMG) obtained using typical motion analysis laboratory methods. Review articles, commentaries, protocols, and short conference abstracts were excluded.

Initial screening for inclusion/exclusion criteria was performed by 1 of 5 experienced gait laboratory personnel. Secondary screening was then performed by at least 2 evaluators to identify the level of efficacy, if any, addressed by each paper. Levels of efficacy were defined similar to our previous publication: level 1 (technical), level 2 (diagnostic accuracy), level 3-4 (diagnostic thinking and treatment), level 5 (patient outcome), level 6 (societal). An additional level (2b) identified papers on the effects of treatments at a group level.

RESULTS

Overall results are presented in Fig. 1. The level 1 studies primarily dealt with development of improved technology for data collection and modeling, including reliability and validity studies. The level 2 studies generally compared different diagnostic or demographic groups, related gait measures to other gait measures, participant characteristics, or testing conditions, or developed methods to advance data interpretation. The level 2b studies mainly evaluated the effects of treatments such as surgery, bracing, prosthetics, or rehabilitation programs or compared the outcomes of different treatments.

The level 3-4 studies showed that 3DGA changes treatment plans and increases clinicians' confidence in their treatment decisions for patients with cerebral palsy (CP) [2], spina bifida [3], and post-stroke [4]. There is little agreement in problem identification and surgical planning between clinical examination and 3DGA [5]. However, agreement among clinicians in problem identification and goal setting increases with 3DGA [6].

The level 5 articles included two from a randomized controlled trial (RCT). The first [8] showed little difference in 1-year outcomes between children with CP randomized to receive a pre-operative 3DGA report and controls who underwent surgery without the report. The lack of difference was attributed to low adherence to the 3DGA recommendations, which was only 42% in the group receiving the report compared with 35% in the control group [7]. This is much lower than the 77-97% adherence typical of clinical referrals [8]. To investigate this effect, a second paper subdivided the group receiving the gait report according to whether or not a particular recommendation for femoral derotation osteotomy was followed and found significant improvements into the normal range only when the 3DGA recommendations were both received and followed [9]. A final level 5 study [10] found that the incidence of severe crouch gait dropped impressively, from 25% to 4%, in the years following practice changes including the addition of 3DGA and single event multilevel surgery, which requires 3DGA.

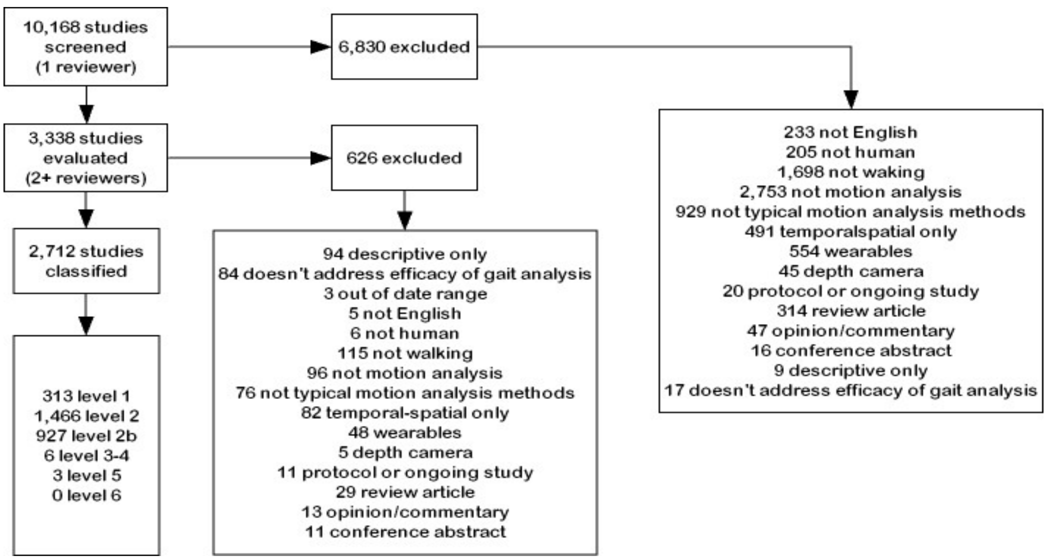

Figure 1: Review flowchart

DISCUSSION

The volume of studies on 3DGA has exploded over the last decade. Thousands of articles contribute to continued development of methods for improved data collection and interpretation. Six new studies in level 3-4, including a RCT, support the findings of 11 similar studies from the previous review period, clearly demonstrating the efficacy of 3DGA in changing and reinforcing treatment decisions, increasing clinicians' confidence in treatment planning, and increasing agreement among different clinicians. Three new studies in level 5, including two from a RCT, demonstrate the potential of 3DGA to improve patient outcomes.

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THE GAIT OUTOMES ASSESSMENT LIST (GOAL) QUESTIONNAIRE: CONSISTENT MEASUREMENT OF FUNCTION ACROSS GAIT CENTERS

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INTRODUCTION: Parent- or self-reported functional outcome measures are important data collected during a typical 3-D gait study to capture patient performance and function in the community. They guide clinicians to information regarding pain, tripping and falling, and level of assistance required to negotiate terrain and distance, but are typically limited in gaining insight into goals and expectations for treatment [1-3]. The Gait Outcomes Assessment List (GOAL) Questionnaire is new assessment of function for ambulatory children with cerebral palsy (CP) designed to address the gap in understanding of patient and family priorities for gait and intervention [4]. Validation was completed in Australia, but has not been widely used in other centers [5]. The purpose of this study was to determine if the initial findings of the GOAL validation can be replicated at another gait center servicing a high population of children with CP.

CLINICAL SIGNIFICANCE: Outcomes from gait analysis-guided clinical decision-making may be improved when family priorities and expectations are understood and incorporated into treatment planning. Use of an outcome measure designed to provide this information must be reliable across multiple centers.

METHODS: Parent version 5.0 of the GOAL questionnaire was initiated into routine clinical use in our center in June 2018. Parents of all patients with a diagnosis of CP completed the GOAL as part of a battery of functional assessments administered during a standard of care 3-D gait analysis assessment. Standardized item, domain and total scores were calculated for each individual. Gillette Functional Assessment Questionnaire (FAQ) walking scale value and Gait Deviation Index (GDI) were recorded/calculated per standard protocol and used for validation assessment. Individuals with a complete dataset (all domain and total GOAL scores able to be calculated) between June 2018-December 2019 were included in the analysis. Item data were assessed to identify the most important goals for families across the domains. Statistical analysis was performed using XLStat and Microsoft Excel 2010. A one-way ANOVA was used to assess differences in total GOAL and domain scores by Gross Motor Function Classification System (GMFCS) level. Spearman's rank correlation was used to assess correlation between FAQ walking level and total GOAL scores. Results were compared to published values [5].

RESULTS: Data from 161 individuals (90 M, 71 F, mean age 11.2±6.1 years) were included. GMFCS breakdown was as follows: I: 38, II: 76, III: 47. Table 1 demonstrates the summary statistics for total and domain scores with comparison to previously published data. Similar to previous data, total GOAL scores were normally distributed. A one-way ANOVA found a significant difference between total GOAL scores by GMFCS level $p<.001$. Figure 1. Positive moderate correlations were noted between total GOAL scores and FAQ and GDI (data not included). Item analysis identified most frequent priorities for change were in the Pain Discomfort & Fatigue, and Gait, Pattern & Appearance domains.

Table 1: GOAL Domain/Total Score Comparison Across Centers

	Gillette Data	Published Data [5]
	mean (SD)	**mean (SD)**
Domain A: Activities of Daily Living & Independence	67.9 (21.2)	75.0 (21.7)
Domain B: Gait Function & Mobility	58.3(21.9)	65.8 (20.9)
Domain C: Pain Discomfort & Fatigue	76.6 (19.1)	73.2 (20.9)
Domain D: Physical Activities Sports & Recreation	30.6 (19.7)	43.1(22.4)
Domain E: Gait Pattern & Appearance	44.0 (21)	46.3 (24.1)
Domain F: Use Of Braces & Mobility Aids	48.8 (27.1)	50.8 (27.7)
Domain G: Body Image & Self Esteem	50.7 (18.9)	47.4 (19.1)
Total Goal Score	**55.4 (14)**	**59.9 (16.9)**

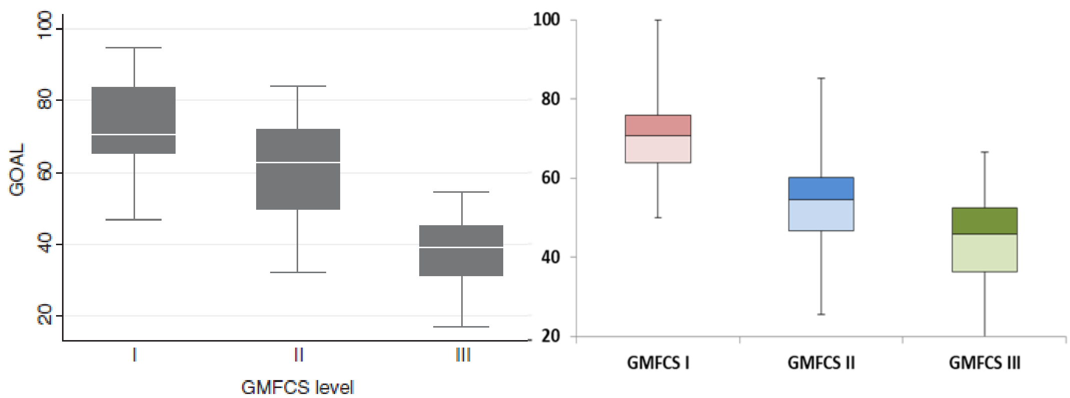

Figure 1: Comparison of published data (left) to Current Center data (right). Box plots represent the 25-75th percentile. Error bars delineate maximum and minimum values.

DISCUSSION: Function assessment using the GOAL questionnaire among individuals with CP demonstrates can be replicated across centers. This increases the confidence that the tool is able to discriminate function among GMFCS levels and is a useful adjunct to care of the ambulant child with CP. The identification of family priorities most commonly in domains C and E also contributes to confidence in the tool. A means to document family priorities for treatment is now available. Future work will assess the tool's sensitivity to change after intervention.

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DISCLOSURE STATEMENT

Authors have no conflicts of interest to disclose.

A MULTI-TASK, MULTI-CENTER MOTION ANALYSIS PROTOCOL: RELIABILITY ASSESSMENT IN HEALTHY INDIVIDUALS

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INTRODUCTION

An important factor in multi-center collaborations for motion capture studies is the assessment of intra- and inter-tester reliability for each component of the study outcome assessments. Although multiple motion capture systems are available commercially, technical standards (accuracy and resolution of cameras, for example) tend to be far within the bounds of the “human factor” involved in motion capture (marker placement error, skin motion artifact, differences in task instruction, intra-participant variability, etc.)

CLINICAL SIGNIFICANCE

The aim of this study was to assess inter-tester reliability in the execution of a standardized, prospective research protocol to quantitatively assess movement patterns for a future multi-center clinical trial in patients with hip pathology.

METHODS

In order to assess the variability in the ***execution*** of the motion analysis protocol, researchers from five institutions individually traveled to a single institution for a 3-day period to perform testing on the same participants. Hardware remained consistent for all testing and was calibrated daily using that institution’s standard protocols. All data were processed by a single research team member using the same modeling software. Extensive pre-testing protocol development and planning, which included multiple conference calls and three in-person meetings were used to develop, define and review, all aspects of the protocol. A detailed standard operating procedure (SOP) was developed and reviewed prior to, and accessible at the time of, data collection for each of the researchers from all institutions. This SOP consisted of detailed instructions for all clinical measures, marker placement, and specific patient setup, verbal instructions and minimal/specific performance criteria needed for trials to be considered “successful” for each functional task were included in the protocol.

Five healthy adult participants were tested by a researcher from one the five institutions during a two- to three- day window. The participants then return for testing 4 more times (once per researcher from each of the other institution) over a three-month period. In addition, one of the participants was tested twice by each researcher. Kinematic and kinetic data were collected during a) static poses, b) level, overground gait at a self-selected speed, c) bilateral stance deep squatting, and d) step-down. Temporal and kinematic variables of interest were determined for each task, consisting of averages, minimums and maximums of relevant joint movements.

Variability was assessed a) across multiple trials within each participant testing session (intra-session) b) within each researcher as a comparison of two test sessions for one participant (intra-researcher) and c) across researchers for each of the five participants (inter-researcher). Minimal detectable change (MDC) was determined to give an overall indication of repeatability.

RESULTS

Across all tasks transverse plane variables, specifically hip rotation, had the greatest MDC.

During gait, temporal parameters were well within published ranges and joint kinematics were within 5° (with exception of hip rotation).

The squat depth (SD) varied by 10%. Generally intra-session variability was highest for the squat, which lead to slightly higher MDC values for some variables across all planes.

Variables during the step-down task were assessed at 60° of knee flexion. Results showed MDC in most joint kinematics were within 8°, with the exception of hip rotation (10.2°). Follow-up discussion and clarification were used to identify issues in the speed in which the squat and step-down task were performed as these were not controlled for in the protocol.

Table 1: Select variables for a) gait, b) squatting and c) stepdown. Values present reliability in degrees unless otherwise stated.

a) Gait Variables	Intra-Session	Intra-Researcher	Inter-Researcher	MDC
Stride Length (m)	0.10	0.14	0.07	0.09
Cadence (steps/min)	3.5	5.5	3.9	4.8
Velocity (m/s)	0.10	0.20	0.10	0.12
Avg Trunk Lean (°)	1.2	1.6	1.2	1.5
Pelvic Tilt ROM (°)	1.5	4.5	3.7	4.5
Hip Flex/Ext ROM (°)	3.7	4.3	2.5	3.1
Avg Hip Rotation (°)	1.8	10.1	9.8	12.2
Avg Knee Var/Valg (°)	0.3	2.4	2.1	2.5
Knee Var/Valg ROM (°)	1.5	4.1	3.4	4.2

b) Squat Variables at Squat Depth	Intra-Session	Intra-Researcher	Inter-Researcher	MDC
% Squat Depth	3.0	14.8	8.2	10.2
Trunk Tilt (°)	4.2	11.2	6.8	8.4
Pelvic Obliquity (°)	5.7	4.3	3.9	4.9
Hip Rotation (°)	6.4	10.8	8.1	10.1
Ankle Dorsiflexion (°)	2.2	3.6	2.7	3.4

c) Step-Down Variables at 60° of knee flexion	Intra-Session	Intra-Researcher	Inter-Researcher	MDC
Pelvic Obliquity (°)	3.0	3.2	3.4	4.1
Hip Adduction (°)	3.5	7.0	4.1	5.1
Hip Rotation (°)	5.1	8.5	8.3	10.2
Knee Flexion (°)	1.9	0.4	2.4	2.9
Knee Var/Valg (°)	1.0	5.3	6.4	7.9

DISCUSSION

Despite having written, standard instructions and protocols during testing performed across motion analysis personnel from multiple institutions, most of the variability came from marker placement and the participant variability in executing the requested motion. Additional training and discussion in marker placement, despite the use of experienced researchers, and the importance to the adherence of SOP task instructions were identified as two areas to address which could potentially improve reliability.

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The Effect of Augmented Plantar Feedback on Walk Ratios

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INTRODUCTION

Barefoot and shod gait has been shown to display considerable differences in kinematics, kinetics, and muscular activation [1, 2]. A possible source of these changes is the alteration in plantar feedback during shod gait. For instance, Sacco et al. found changes in plantar sensory information during shod and barefoot walking resulted in alterations in the lower extremity muscle activity in both healthy individuals and those with peripheral neuropathy. While the exact mechanism(s) that cause these differences has been a point debated in the literature, it is apparent the that the successful integration of sensorimotor information is vital to control mechanisms during gait, especially in populations with decreased capabilities (e.g. older adults, pathology).

More recent research has indicated the ability of augmented tactile feedback to significantly alter spatiotemporal gait parameters and gait symmetry measures (e.g., stance phase, single support, and swing phase) in young, healthy adults, while others have reported a reduced stride length and walking velocity when older adults wore a textured as opposed to a smooth insole [4, 5]. Likewise, the research by Nurse et al. (2005) has indicated the ability of augmented tactile feedback to alter gait mechanics and muscle activity during walking.

Given the influence of plantar sensory information on motor control strategies, analysis of spatiotemporal gait patterns utilizing the velocity independent variable, walk ratio (step length (mm) / cadence (steps*min^{-1})), may assist in better understanding motor control changes resulting from diminished plantar feedback while shod [7-8]. Specifically, previous research has found that the walk ratios are altered in populations with neuromuscular dysfunction (e.g. Parkinson's disease), older adults, and during the execution of dual tasks [8-10]. Therefore, the purpose of the present study was to compare the influence of augmented plantar feedback while shod and barefoot on stride length, stance width, and the walk ratios during free walking (i.e. self-chosen). A secondary purpose was to examine potential gender differences in walk ratios which has not been previously reported.

CLINICAL SIGNIFICANCE

Walk ratios have been utilized as an indicator of alterations in motor control strategies during walking [7-8]. Given the potential influence of plantar sensory information, as well as footwear, on motor control strategies, examining walk ratios during gait may provide a rather simple variable to calculate that would assist clinicians in quickly identifying changes in neuromuscular strategies during gait.

METHODS

Fifty healthy participants (25 male, 25 female) walked across an instrumented walkway (GAITRite, CIR Systems, Inc., Havertown, PA, USA) at a normal, self-selected pace during four footwear conditions: barefoot (BF), insole-only (IN), a minimalist running shoe (SH), and a minimalist running shoe with the textured insole (INSH). Spatiotemporal variables were averaged across six trials of normal walking for each of the footwear conditions. The mass of the insole-only, shoe, and insole/ shoe combination was 31.5g, 143g, and 162g, respectively. A two-way (IV's: Footwear, Gender) repeated measures ANOVA's with Bonferonni post hoc analyses were performed with dependence on Walk Ratios and Walking Velocity. Stride length (mm) and cadence (steps*min^{-1}) were collected to calculate the walk ratio (step length (mm) / cadence (steps*min^{-1})) during each condition.

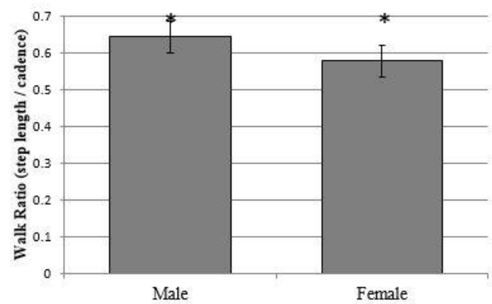

Figure 1: Gender Differnces in Walk Ratios
*Males displayed a significantly greater ($p < 0.001$) walk ratio as compared to females.

RESULTS

Statistical analyses revealed a significant main effect for gender ($p < 0.001$) as well as footwear condition ($p < 0.001$); no significant effect was noted for velocity. Specifically, results indicate that females display significantly smaller walk ratios as compared to males (Figure 1).

With respect to walk ratios (Figure 2), follow-up analyses indicated that participants displayed significantly greater ($p < 0.001$) walk ratios when Shod (SH) as compared to Barefoot (BF) as well as when Shod with a textured Insole (INSH) as compared to BF ($p < 0.001$). Finally, results indicate that there was no significant difference in the walk ratios of individuals with BR as compared to the insole-only condition (IN).

Figure 2: Footwear and insole effects on Walk Ratios
1: Participants displayed a significanly greater ($p < 0.001$) walk ratio during SH as compare to BF or IN
2: Participants displayed a significanly greater ($p < 0.001$) walk ratio during INSH as compare to BF or IN

DISCUSSION

Previous research has found walk ratios to be a reliable measure for identifying changes in motor control strategies during gait [7-10]. Furthermore, research has previously reported significant differences in barefoot and shod gait, including altered efferent muscular activation patterns, as well as the ability of augmented plantar feedback to alter gait patterns in a variety of populations [1-6]. The present study suggests that while footwear does indeed alter the walk ratios of individuals, augmented plantar feedback has a minimal effect on gross motor control strategies. Furthermore, it has been previously reported that a smaller walk ratio may be an indicator of potential pathology; however, the present study suggests that females with no pathology will often display significantly lower walk ratios as compared to males [7-8]. While not altogether unexpected given the smaller stature of females, this finding does indicate the potential need for gender specific norms.

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COLLECTION AND PUBLICATION OF ADULT NORMATIVE GAIT DATA

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INTRODUCTION

The small sample sizes of public gait databases represent a major problem in this field since it limits any statistical inferences that can be made from these data. When the typical gait study of 10-30 subjects [1] is contrasted with those (non-gait studies) involving thousands of subjects, the gait studies run the risk of not being taken seriously by the larger medical community or insurance industry. Compounding the problem of low numbers is the lack of raw marker data. In general, when gait data is made public, it is in the form of joint kinematics and kinetics (angles, forces, and moments). While Moore et al. [2] describe larger databases of this nature, such databases are of limited use by other researchers seeking to compare their data to such norms unless identical marker sets and kinematics models were used. Given the same motion and marker set, different joint angles, forces, and moments may be obtained by the use of different algorithms. When marker sets are changed, this problem is exacerbated. Even knowing the algorithm and marker set used to create the joint kinematics and kinetics from an external database, it can be close to impossible to reconstruct original marker positions and thereafter apply a lab's own kinematics code to gain normative joint angles that are comparable to those obtained in-house. The only true solution is to start with raw marker data.

CLINICAL SIGNIFICANCE

In orthopedics, normative data are fundamental to the interpretation of patient results. The availability of free and open-sourced raw marker data from a relatively large number of subjects allows for patient to norm comparisons to be made in a more meaningful way between laboratories.

METHODS

After obtaining IRB approval, we recruited subjects between the ages of 18-65 years from the Penn State and surrounding communities as subjects. Subjects who were currently injured or in pain, in need of an assistive device, pregnant, who have had a joint replacement, or who were unable to consent, were excluded from this study. However, subjects who have had prior lower limb surgical intervention for any reason *were* included, and this was noted in the data without specificity to preserve subject anonymity. Subjects wore shorts and tank tops, and were marked with a modified Cleveland Clinic marker set totaling 40 markers. An 8 camera Motion Analysis Corporation system was used to collect data kinematic data at 100 Hz, and subjects were asked to walk at their self-selected speed across the lab 10 times. Subjects also provided their age, height, weight, gender, and whether or not they had had any orthopedic interventions. Body mass index (BMI), forward velocity, height normalized velocity, stride time, stance width, step length and symmetry based on step length were calculated from representative trials. Tracked marker coordinate data along with a summary spreadsheet containing the noted ancillary information were posted to GitHub (https://github.com/timniiler/gaitdata) under the Lesser GPL v2.1 license.

RESULTS

At the time of this writing 37 subjects (28 Male, 9 Female) have been collected at Brandywine. A summary of temporal-spatial and physical characteristics is listed in Table 1 below.

Table 1: Characteristics of data thus far, means (sd) by gender.

Sex	Age (y)	Height (m)	Weight (kg)	Ortho (Y/N)	BMI (kg/m^2)	Mean Vel (m/s)	Norm Vel (s^{-1})	Stride Time (s)	Stance Width (m)	Step Length (m)	Symmetry Ratio
F	42.2 (10.6)	1.65 (0.04)	61.4 (6.1)	3/6	22.7 (2.5)	1.19 (0.23)	0.72 (0.13)	1.03 (0.08)	0.14 (0.04)	0.62 (0.07)	0.91 (0.13)
M	22.9 (11.1)	1.77 (0.08)	78.1 (15.3)	4/23	24.8 (4.4)	1.18 (0.14)	0.66 (0.08)	1.12 (0.08)	0.14 (0.04)	0.67 (0.07)	0.92 (0.06)

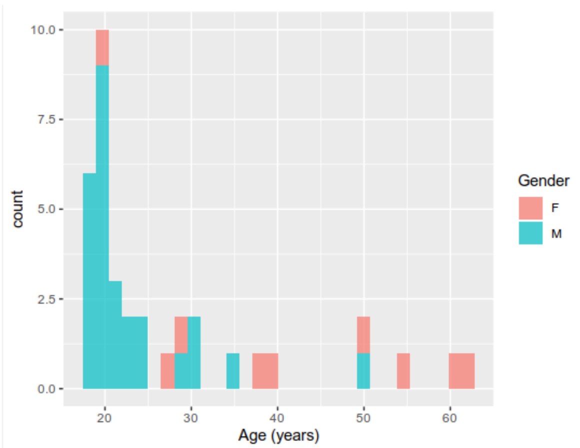

Figure 1: Distribution of subject age by gender

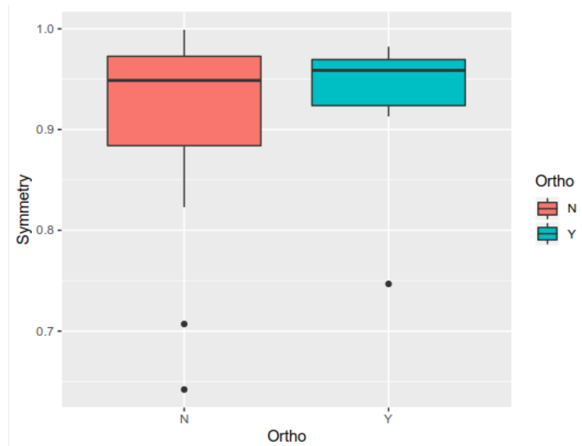

Figure 2: Symmetry by orthopedic intervention

DISCUSSION

This project offers researchers from other laboratories normative data which can be used as a control set for clinical use or for other projects as needed. The physical characteristics of our participants thus far is fairly typical of the adult population at large in the areas of BMI, height, and weight [3] as are the temporal-spatial results [4]. However, the data are skewed towards the younger population due to the location of subject recruitment. One interesting result is that those subjects who had orthopedic interventions were more symmetrical in their step length than those who had not bringing into question precisely what “normative” means. Data collection is ongoing, and there are plans to continue to add subjects, and to renew the IRB proposal when it expires in May 2020.

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DISCLOSURE STATEMENT

The authors have no conflicts of interest to disclose.

CONFLICTING DATA IN THE TRANSVERSE PLANE: ASSESSING THE IMPACT OF SURGICAL DECISION-MAKING

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PATIENT HISTORY

CM is a 6 year 6 month old boy with a diagnosis of spastic diplegia cerebral palsy. He was referred by an outside physician for 3-D gait analysis in consideration of bilateral femoral derotation osteotomies to correct in-toeing, reduce tripping and falling and improve function. CM is approximately 18 months status-post selective dorsal rhizotomy (SDR) utilizing gait analysis guided clinical decision-making in 2016. He functions at a GMFCS level I, and falls multiple times per day. This remains unchanged since prior to his SDR.

CLINICAL DATA

Physical examination included the following bilateral findings: 1) 60-70° femoral anteversion by trochanteric prominence test bilaterally; 2) internal bimalleolar axis (R>L); 3) limited dorsiflexion bilaterally R>L; 4) grade 3-4 strength by manual muscle testing at hip and knee and grade 2-3 at foot and ankle; 5) resolution of spasticity by Ashworth scale; 6) hypermobile midfoot on non-weightbearing foot exam with decreased medial longitudinal arch in weightbearing.

MOTION DATA

Data is collected with a 12-camera motion capture system (120 Hz; Vicon, Oxford UK) using a modified Helen-Hays marker set and functional model calibration[1]. See Figure 1.

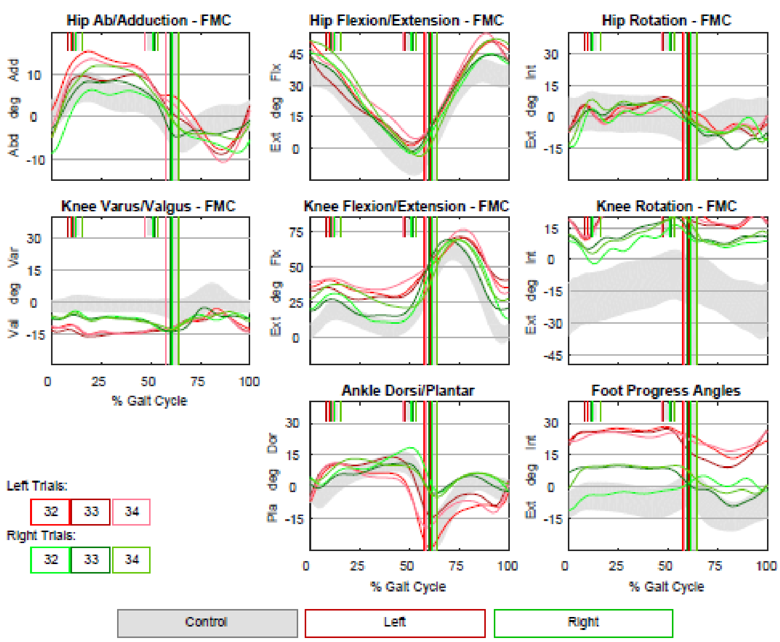

Figure 1: Kinematic consistency data vs control in the coronal (left column), sagittal (center column), and transverse (right column) planes. FMC=functional model calibration. Multiple trials are superimposed on the same graph.

TREATMENT DECISIONS AND INDICATIONS
Data demonstrate the following deviations: 1) clinical femoral anteversion in the absence of excessive internal hip rotation during gait 2) bilateral internal tibial torsion with associated internal foot progression angles; 3) excessive knee flexion bilaterally L>R; 4) bilateral knee valgus in stance; and 5) borderline left foot drop in swing.
Treatment recommendations: 1) bilateral long-leg x-rays to examine for knee valgus 2) CT scan and examination under anesthesia with percutaneous pin to assess magnitude of femoral anteversion; 3) bilateral external tibial derotation osteotomies 4) possible injectable medication to left gastrocnemius; 5) flexible posterior leaf spring bracing on left.
Treatment: 1) bilateral varus femoral derotation osteotomies (10° varus, 30° rotation); 2) bilateral tibial derotation osteotomies (20°).

OUTCOME
Repeat gait data document incomplete correction of tibial torsion, and change from neutral to external hip rotation subsequent to femoral derotation osteotomies bilaterally. Foot progression angles are restored to typical. See Figure 2.

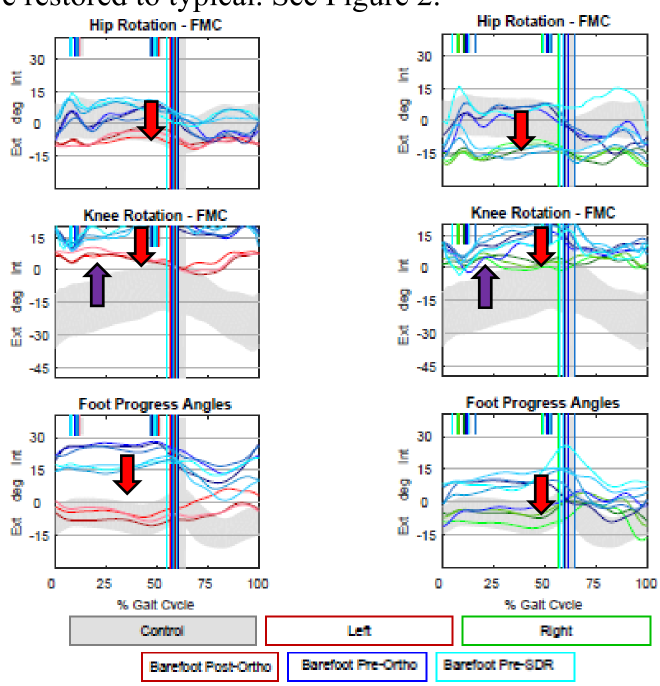

Figure 2: Post-ortho vs pre-ortho vs. pre SDR transverse plane comparisons for left and right sides. FMC=functional model calibration. Red arrows document areas of improvement. Purple arrows highlight persistent internal knee rotation (tibial torsion).

SUMMARY
Tibial derotation osteotomies were not part of original pre-gait study consideration. This case study documents influence of under- and over-correction of deformity. Results however, for patient were neutral foot progression angles with decreased falling.

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

RELIABILITY OF TRUNK AND LOWER EXTREMITY KINEMATICS USING MODIFIED PLUG-IN GAIT AND OXFORD FOOT MODELS DURING TREADMILL RUNNING: A PILOT STUDY

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INTRODUCTION

Over 30 million Americans run competitively or recreationally. Twenty percent run on a regular basis [1,2]. The incidence of injury amongst runners ranges from 20-80% [1-3]. Understanding the three-dimensional (3D) kinematics and kinetics of running may be useful to diagnose or prevent running-related injuries, and enhance performance [4]. However, before instrumented running analysis can be used clinically a control data set, and its reliability must be examined. The Plug in Gait (PiG) and Oxford Foot (OFM) models have independently demonstrated face validity and reliability for biomechanical variables associated with walking gait in adults and children [5,6]. However, there is paucity of evidence on reliability of kinematic data when these models are used during shod running. The purpose of this study was to examine inter-examiner reliability of select trunk, pelvis and lower extremity kinematic variables in healthy adults using PiG and OFM during shod treadmill running.

CLINICAL SIGNIFICANCE

Three-dimensional kinematic analysis of running biomechanics may be useful to assist in performance enhancement and injury management. Use of a multi-segment foot model may be especially useful to provide insight into running injuries. However, established valid and reliable control data sets that include use of multi-segment foot models are not available. This project provides a necessary first step.

METHOD

This study was approved by the Institutional Review Board at Grand Valley State University (17-222-H). Twenty healthy recreational runners (running at least 15 miles/wk) aged 18 to 35 years (27.9 ± 8.7 yrs; 8 males, BMI 25.6 ± 1.4 kg/m^2; 12 females, BMI 20.9 ± 2.5 kg/m^2) participated. Ten runners (6 female, 4 male) returned approximately one week later for retesting. Fifteen Vicon (Oxford Metrics, UK) cameras (120 Hz) were synchronized with an AMTI instrumented (1200 Hz) treadmill (Advanced Mechanical Technology, Inc, Watertown, MA, USA). Markers were placed by two trained doctoral physical therapy students. With the Oxford Foot model markers were placed directly on the foot through pre-cut holes in standardized running shoes (Mizuno Wave Runners). Data were collected at three speeds (6.21mph, 7.45mph, and 8.60mph), reduced using Vicon Nexus, v 2.8.1., and exported to Visual 3D (C-Motion, Inc., Germantown, MD) to determine joint kinematics and kinetics. Three right and three left representative gait cycles from each participant were analyzed. Statistical analysis, using JMP® 13 (SAS Institute Inc., Cary, NC) was completed on kinematic data for only the 7.45 mph running speed. A variance components mixed model analysis was utilized to assess the variance associated with intrinsic and extrinsic sources of error, with total therapist session variation representing inter-rater reliability, with alpha set at <0.05 [7,8].

RESULTS & DISCUSSION

We considered total session variability, i.e., standard deviation (SD), less than 2º as good, between 2º-5º as acceptable, and above 5º as poor reliability [9]. Total therapist variation across all joint angles ranged from 0.8º-7.8 º (Table). Kinematic variability due to therapist was greatest for sagittal plane trunk/pelvis, pelvis, and hip, as well as transverse plane hip and knee; and largest in all planes forefoot/hindfoot, and hallux/forefoot. Twenty-four out of 28 segments/joints had a significant Wald p-value in at least one total therapist variation component. Kinematic variability tended to be greater for foot kinematics. We believe that most variability was related to soft tissue artifact and inconsistencies in marker placement. Our data demonstrate reasonable to good inter-therapist reliability, and that placing markers on the foot inside shoe-cut holes could be utilized for future investigations. We have also demonstrated that a variance components analysis is feasible and clinically useful in studying reliability of 3D gait. It is our recommendation that this statistical model be used in future studies instead of coefficient of multiple correlation (CMC) and intraclass correlation coefficient (ICC) to determine and interpret reliability of 3D gait. Finally, our data suggest that the PiG and Oxford foot models could be used for longitudinal evaluation of runners.

Table. Standard deviation (SD)* of total session variation reported in degrees.

PiG Segments/Joints		Trunk			Trunk - pelvis			Pelvis			Hip			Knee		
Plane		Sagittal	Frontal	Transverse	Sagittal	Frontal	Transverse	Sagittal	Frontal	Transverse	Sagittal	Frontal	Transverse	Sagittal	Frontal	Transverse
Total Therapist Variation	SD	0.8	1.0	2.7	2.7	1.2	2.1	3.1	1.4	1.4	4.6	1.8	4.4	3.5	2.9	5.0
	% Tot Var	3.6	13.5	35.8	21.1	8.2	18.2	48.0	23.0	13.8	33.6	22.7	16.3	11.0	10.4	27.0
Total Variation	SD	2.8	2.2	3.6	5.6	3.7	3.9	3.8	2.5	3.1	6.0	3.7	8.0	8.2	6.4	7.4

OFM Segments/Joints		Ankle		Foot Progression	Forefoot - tibia			Forefoot - hindfoot			Hindfoot - tibia			Hallux - forefoot
Plane		Sagittal	Frontal	Transverse	Sagittal	Frontal	Transverse	Sagittal	Frontal	Transverse	Sagittal	Frontal	Transverse	Sagittal
Total Therapist Variation	SD	2.7	1.4	1.2	5.3	3.0	4.4	7.8	6.6	4.1	4.5	4.4	4.9	5.6
	% Tot Var	5.5	10.2	3.7	13.4	26.4	24.4	60.5	59.1	44.0	25.6	67.0	25.0	31.7
Total Variation	SD	7.3	4.4	6.4	8.6	3.8	6.6	6.0	5.8	5.0	6.8	4.6	6.2	6.2

*Calculated by combining the three effects containing session data; percent total variance (% total variance) of total session variation, calculated by summing % total variance of session effects; and SD of total variation, calculated from the combination of all six effects. Bold indicates an effect Wald p-value < 0.05, with total therapist variation bolded if at least one therapist-related effect had a Wald p-value < 0.05.

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MEASURES OF DYNAMIC BALANCE DURING AMBULATION UNDER SINGLE- and DUAL-TASK CONDITIONS IN FOOTBALL PLAYERS: A PILOT STUDY

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INTRODUCTION

The prevalence of sports-related concussion is purported to range from 1.6-3.8 million per year, with perhaps 50-80% unreported incidences [1,2]. Sports-related concussions result in subacute and chronic deficits and functional limitations that include altered cognition and postural control, impaired gait, difficulty with dual task activities, and increased risk for other musculoskeletal injuries [3]. There does not appear to be standard return-to-play (RTP) protocols, although many utilize the Sport Concussion Assessment Tool that includes a modified Balance Error Scoring System (BESS) [4]. However, it appears that current RTP protocols include tests that are not sensitive enough to detect subtle central nervous system damage resulting in many athletes premature return to sports [5]. Recent research has suggested that gait (under both single- and dual-task conditions) and other dynamic balance measures, which include the use of more complex biomechanical variables, e.g., center of pressure (COP), COP velocity, etc., provide more critical evaluative criteria for concussed athletes [3]. Albertson et al. [6] have examined differences in COP displacement and anterior/posterior (AP) and medial/lateral (ML) velocity in healthy athletes under dual-task conditions, but no one has examined differences in these variables over time and condition. And Grants et al. [7] have found differences in performance under gait dual-task conditions in concussed athletes. However, there are no longitudinal single- and dual-task gait studies that have examined differences in COP variables in non-concussed football players. Therefore, the purpose of this study was to examine for differences in dynamic balance variables under single and dual-task conditions during self-paced walking one year apart in non-concussed football players.

CLINICAL SIGNIFICANCE

Awareness of variation in dynamic balance indices related to specific sub-phases of gait over time for contact and collision sport athletes is important. These indices could be used to in RTP protocols and assist with identifying differences between concussed and non-concussed athletes.

METHOD

Seven healthy, non-concussed Division II collegiate football players (18.6$\pm$0.5 yrs.; height: 188.0$\pm$5.5 cm; mass: 115.0$\pm$22.0 kg; BMI: 32.3+4.9 kg/m^2) participated. Ethics approval was obtained from the Grand Valley State University Human Research Review Committee (IRB# 1010798-1). Ground reaction force (GRF) data were collected using AMTI NetForce software (Advanced Mechanical Technology, Inc., Watertown, MA) as participants walked at a self-selected pace across a portable AMTI force platform (200 Hz) under single- (ST) and dual-task (DT) conditions. Cognitive tasks during DT included spelling 5-letter words backwards, subtracting 3's or 7's, and reciting the months of the years backwards. Five clean force plate strikes for each limb under each condition for year one and two were analyzed. Temporo-spatial gait parameters were determined using G-Walk (BTS Bioengineering Corp., Quincy, MA). A custom MATLAB program was used to extract raw GRF data to dependent variables: maximum medial-lateral (M/L) center of pressure (COP) excursion, mean M/L and anterior-posterior (A/P) COP velocity during first double limb support (DS) and single limb support (SS), and temporo-spatial gait variables throughout gait. SAS JMP (SAS, Cary, NC) and a mixed model ANOVA was used to analyze data with $p < 0.05$ to determine significance.

RESULTS & DISCUSSION

Compared to ST, cadence, stride length (1.45 ± 0.21 m for ST vs. 1.38 ± 0.22 m for DT) and gait velocity (1.31 ± 0.18 m/s for ST vs. 1.19 ± 0.19 m/s for DT) were reduced under DT conditions ($p < 0.005$ for all), consistent with that previously been reported [3].

Maximum M/L COP excursion was significantly greater in single limb support (16.26 ± 0.41 mm for SS; 14.48 ± 0.43 mm for DS, $p = 0.0027$). Greater M/L COP excursion in SS has been associated with less stability during that phase of gait, suggesting that this metric may be important to monitor following concussion [8]. Both mean A/P and M/L COP velocities showed significantly different interactions between year and support ($p < 0.01$) as well as main effects ($p < 0.001$) for both year and support. Only mean A/P COP velocity among three variables differed with task; it increased during ST to 535.9 ± 9.34 mm/s from 492.0 ± 9.66 mm/s at DT ($p = 0.0011$), which corroborates Grant et al. [7]. Figures 1-3 show the mean ± SD of three variables on 8 combined conditions (Task: ST vs. DT, Limb support: DS vs. SS, Year: 1 vs. 2).

It was not surprising A/P COP velocity was greater during the ST condition since it was noted that both stride length gait velocity were also greater in ST. Decreased A/P COP velocity in SS however may be an indicator of less stance stability, especially when paired with greater M/L COP excursion. Current RTP concussion protocols typically use tests of dynamic control, e.g., Balance Error Scoring System (BESS) and timed tandem gait, which may not be sensitive enough to determine subtle central nervous system residual injury and may result in premature sport return. Our data suggest the M/L COP excursion and both M/L and A/P COP velocity may be more sensitive and useful dynamic indices of stability balance during gait.

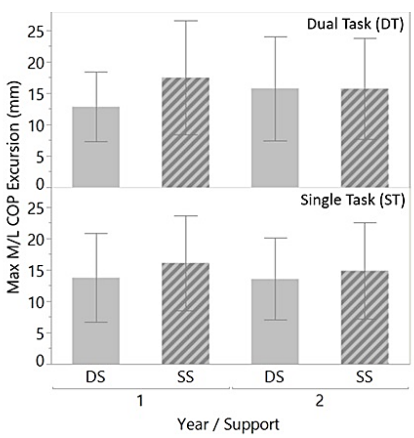

Figure 1. Maximum M/L COP excursion.

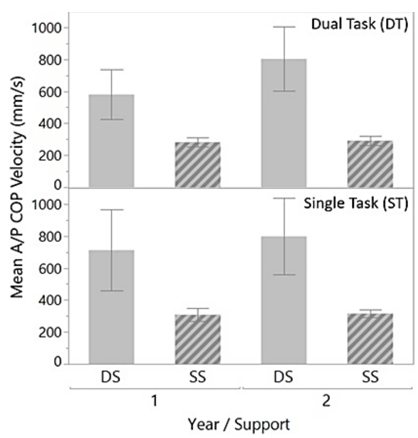

Figure 2. Mean A/P COP velocity.

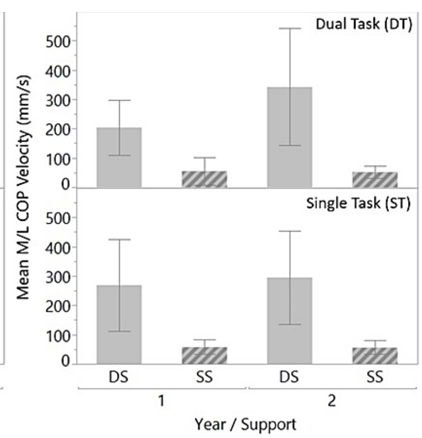

Figure 3. Mean M/L COP velocity.

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Femoral Shaft Gunshot Fractures: Long-Term Post-Operative Gait and Strength

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INTRODUCTION

Gun violence is a national epidemic in the United States, leading to nearly 500,000 victims and more than 50,000 nonfatal injuries annually [1]. The femur is the most commonly fractured long bone associated with gunshot wounds (GSW) [2]. Despite high rates of union and minimal complications, patients often experience long-term pain and discomfort [3]. There is currently little data on the medium-to-long term functional outcomes of post-operative femoral shaft fractures and even less data on femoral shaft fractures due to GSWs.

CLINICAL SIGNIFICANCE

This study analyzed the gait and bilateral leg strength of participants at least 2 years after intramedullary nailing of an isolated femoral shaft fracture due to GSWs in order to provide preliminary data on the functional outcomes of this population.

METHODS

Five males (34 ± 4 years of age) post-operative (5 ± 2 years) from isolated femoral shaft fractures due to GSWs consented to participate in an IRB-approved study protocol. Seventeen reflective markers were placed over key lower extremity anatomical landmarks in accordance with the Plug-in Gait model (Vicon Motion Systems, Ltd., Oxford, UK). Participants were instructed to walk along a 30 ft walkway at a self-selected, comfortable pace until ten trials with consecutive bilateral force plate strikes were recorded. Isometric knee flexion/extension strength was assessed using a Biodex System 3 Pro dynamometer (Biodex Medical Systems, Inc., Shirley, NY). Testing was performed at 30, 60 and 90 deg of knee flexion and compared between the surgically repaired and unaffected limbs. Temporal-spatial gait parameters, kinematics, and kinetics were compared between the population and historical control data. Pointwise statistical analyses were conducted across the gait cycle using Welch's two-tailed, unpaired t-tests and Cohen's d effect size statistical test, with significance levels set at 0.05 and 0.8 (large effect), respectively.

RESULTS

The GSW population displayed diminished isometric quadriceps strength in their affected limb at 60° ($p=0.01$) and 90° ($p=0.01$) of knee flexion and diminished isometric hamstrings strength at 90° ($p=0.01$) of knee flexion. The study population exhibited significantly reduced ($p<0.001$) walking speed (0.82 m/sec vs 1.21 m/sec), stride length (1.03 m vs 1.30 m), and cadence (94 steps/min vs 112 steps/min), as well as a prolonged stance phase (63.3% vs 60.6%). Compared to control data, the GSW population displayed statistically significant ($p<0.002$, $d>1.0$) differences in sagittal plane kinematics and kinetics. At the hip, they experienced reduced dynamic range with absent hip extension, diminished flexor moment in late stance, and decreased peak hip push off power (Figure 1). At the knee, they exhibited reduced maximum flexion angle, decreased dynamic range, and diminished extensor moment demands. At the ankle, they demonstrated reduced dorsiflexion, decreased range, diminshed plantarflexion moment in mid-stance, and reduced total ankle push off power.

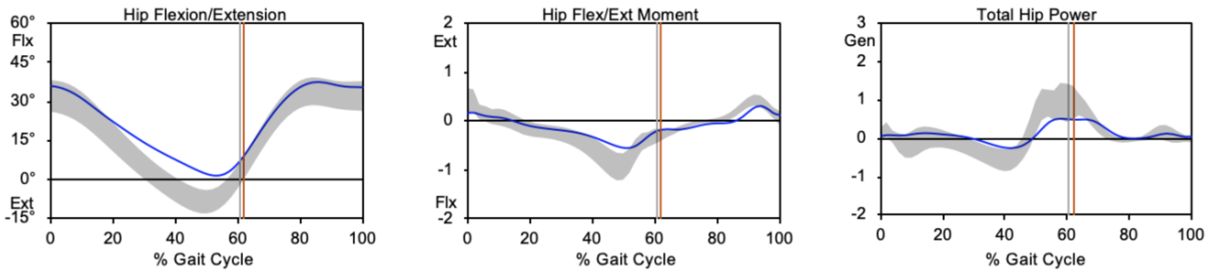

Figure 1: Hip kinematics and kinetics. The solid blue line is the fracture population. The gray band is the average of the control data ± 1 standard deviation. Vertical lines represent toe off for control (gray) and GSW (red) data.

DISCUSSION

The GSW population exhibited significant asymmetrical weakness between their surgically repaired and unaffected limbs as well as gait deviations. The hamstrings and quadriceps had reduced strength, indicating a lack of complete recovery post-operatively. Kinematic and kinetic differences were seen at the hip, knee, and ankle, many quite similar to those seen in elderly populations [4]. Quadriceps dysfunction after injury often lead to reduced knee flexion angle and moment during stance phase [5]. Decreased stride lengths and gait speed are often adopted as part of an adaptive, injury prevention mechanism [5,6]. However, strength imbalances of the quadriceps and hamstrings, as well as reduced stride lengths and gait speeds, have shown to be predictive for future injuries and disability in several populations [5,6]. This work suggests that GSW population could undergo continued rehabilitation to further reduce these disparities to promote their long-term musculoskeletal health. Future work should include evaluation of muscle activation to determine the impact of neuromuscular coordination abnormalities to help guide specific treatment protocols.

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DISCLOSURE STATEMENT

G. Schmeling is on the editorial board of The American Journal of Orthopaedics, The Journal of Orthopaedic Trauma, and The Journal of Surgical Education. All other authors have no conflicts of interest to disclose.

Validation of OnBaseU Clinical Movement Assessment with Biomechanical Motion Analysis in Youth Baseball Pitchers.

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Introduction: Youth baseball pitching injury remains a clinical issue.[1] While overuse is an accepted risk factor for injury, improper pitching mechanics also plays a large role.[2] The current gold standard for evaluating pitching mechanics is a motion capture system. However, most athletes do not have access to these due to their cost, scarcity, and the need for a team biomechanist. The OnBaseU screen is a clinical assessment tool developed to identify movement patterns key to pitching mechanics. If accurate, this tool would serve as a low-cost and portable test that could help achieve efficient mechanics by identifying and tracking movement patterns. However, since it is not yet validated with biomechanical kinematic data, it is unknown if it correctly evaluates the movements it was designed to assess. In this study, we compare portions of the OnBaseU screen with corresponding kinematics using a motion capture system. This analysis will provide validation for this assessment so it may be used as a standard measurement of performance and injury risk specifically for baseball pitchers.

Clinical Significance: If validated, the OnBaseU clinical assessment screen for pitchers could be used to reduce injury risk by identifying harmful pitching mechanics.

Methods: The OnBaseU and motion capture pitching evaluations were completed for 102 youth pitchers (age = 15.2 ± 1.29 years; height = 70.9 ± 3.41 in; weight = 168 ± 30.4 lbs) on the same day. OnBaseU testing was performed by a strength & conditioning coach certified by the College Strength and Conditioning Coaches association (CSCCa) and OnBase University with over 12 years of experience. Motion data were collected at 250 Hz using the 38 reflective marker set required for PitchTrak (Motion Analysis Corporation, Santa Rosa, California) and a ten-camera motion analysis system (Motion Analysis Corporation, Santa Rosa, California). Pitches were thrown at regulation distance from a mound that met major league specifications. Participants conducted a normal warmup, followed by 4 fastballs, 4 changeups, and 4 breaking balls. Kinematic data was obtained for 3 fastballs using PitchTrak.

Four comparisons of motion capture kinematics vs. OnBaseU tests were analyzed: (1) hip-shoulder separation at foot-strike (°) vs. seated trunk rotation test, (2) stride length (%body-height) vs. side step walkout test, (3) stride length (%body-height) vs. push-off test, and (4) shoulder maximum external rotation (°) vs. shoulder 90/90 test. For quantitative analysis, evaluations were assigned points where best to worst performance was awarded the most to least points, which were expressed graphically as “excellent,” “good,” “moderate,” “fair,” and “poor”. A linear regression was fit to each comparison, and R^2 values were calculated. Visual inspection of box and whisker plots showed that the data were not normally distributed, so a non-parametric Wilcoxon two-sample test was performed using JMP (JMP, Cary, NC) to test the difference between group averages.

Results: Motion capture metrics and OnBaseU screens from 102 pitchers were compared. **Figure 1A-D** shows comparisons between motion capture metrics (y-axis) and OnBaseU tests (x-axis). The R^2 statistics from linear regression analysis are as follows: (**A**) hip-shoulder separation at footstrike (°) vs. seated trunk rotation test = 0.036, (**B**) stride length (%body-height) vs. side step walkout test = 0.118, (**C**) stride length (%body-height) vs. push-off test = 0.033, and (**D**) shoulder maximum external rotation (°) vs. shoulder 90/90 test = 0.018. P-values from Wilcoxon non-parametric tests are shown above statistically signficant ($p<0.05$) comparisons in Figure **1A-D**.

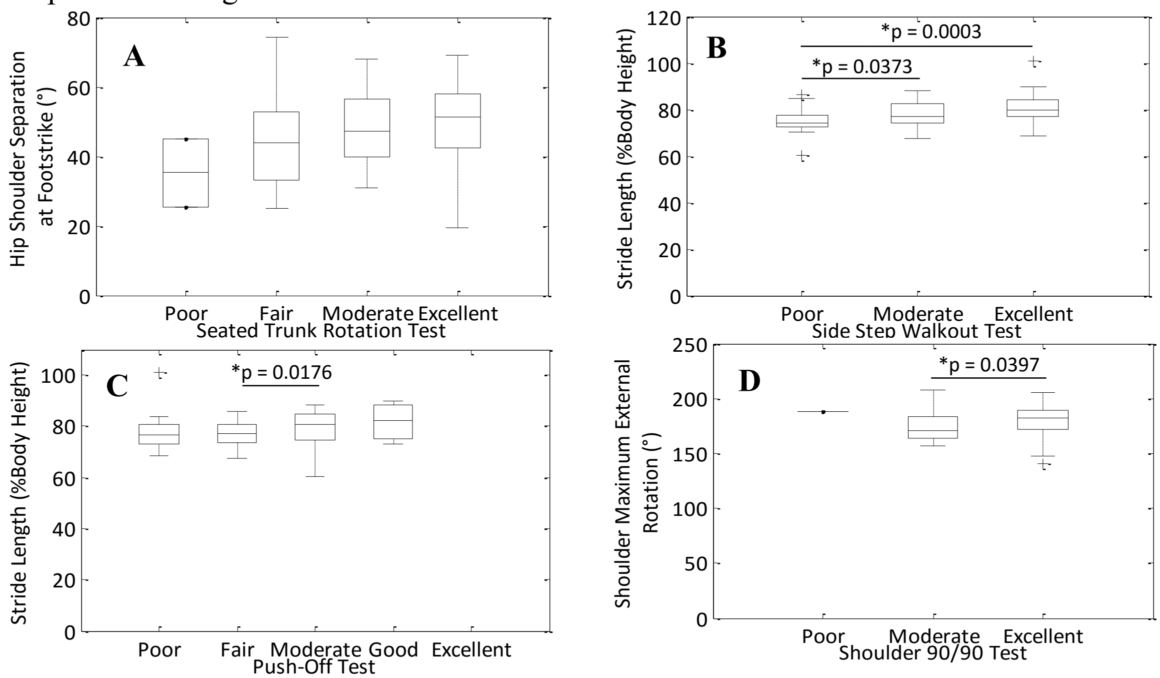

Figure 1. Box and whisker plots display data range. The box shows the 25th to 75th percentile, the line across the interior is the median, and the whiskers display the minimum and maximum. The "+" symbol labels outliers. (A) Comparison of hip shoulder separation at footstrike (°) and the seated trunk rotation test. (B) Comparison of stride length (% body height) and the side step walkout test. (C) Comparison of stride length (% body height) and the push-off test. (D) Comparison of shoulder maximum external rotation (°) and the shoulder 90/90 test. P-values from Wilcoxon non-parametric tests are shown above statistically signficant (p<0.05) comparisons.

Discussion: Linear regressions showed no correlation between motion capture metrics and OnBaseU scores. This may be due to overlap between screening categories or the small number of potential dependent variable results for each test. To determine if the overlap is truly characteristic of the OnBaseU screen, a non-parametric Wilcoxon two-sample test was performed and showed statistically significant results from few groups (**Figure 1A-D**). Most notably, average hip shoulder separation at footstrike (°) and average stride length (%body-height) metrics from players who scored "poor" vs. "excellent" were not statistically signficantly different, indicating overlap of categories that are on opposite ends of the OnBaseU screen scoring spectrum. The lack of statistically significant differences between groups shows that there does exist overlap between screening categories, indicating that players may receive different scores ranging from "poor" to "excellent" in the OnBaseU screen but still demonstrate similar pitching mechanics. These results suggest that the OnBaseU screen lacks the sensitivity to predict pitching mechanics.

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GAIT ANALYSIS AND KINEMATICS OF KNEE JOINT IN PATIENTS IN ACUTE PHASE ANTERIOR CRUCIATE LIGAMENT TEAR

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INTRODUCTION

Anterior cruciate ligament (ACL) injury remains a common injury of the knee joint [1;2]. ACL tear accounts for a considerable part in the overall epidemiology of knee joint (KJ) injuries and may account for 50% of all cases [3].

Classification of ACL injuries based on the time post-trauma also varies. There are no clear criteria for time-based classification of ACL injuries [4].

The most typical functional symptoms in the acute period of ACL injury include asymmetric gait pattern with reduced knee joint flexion and extension ranges of motion in the stance phase [5] of the gait cycle (GC).

CLINICAL SIGNIFICANCE

The objective of the study was to identify the most typical functional symptoms of acute ACL injuries.

METHODS

The control group included 20 healthy adults: 14 men and 6 women. The mean age was 29.7 years. 18 patients with ACL tear were examined 3 to 38 days post-trauma, with the mean post-trauma time being 15.9 days. For 10 of 18 patients the time from trauma to examination was up to 2 weeks. A total of 11 male and 7 female patients were examined, their mean age being 34.1 years. Based on the results of examination, the patients were divided into two groups: with severe function impairment (6 patients) and with moderate function impairment (12 patients).

Inertial sensors placed to the pelvis, the thigh and the shank of both sides studied the gait analysis. The following temporal parameters were determined: GC time (c); the beginning of second double support phase (SDS - % of GC). Movements in the joints were analyzed as follows: for hip and knee flexion-extension, we recorded maximum flexion amplitude in the weight acceptance period, maximum extension amplitude. The obtained data were processed using standard ANOVA methods of the Statistica 10 package.

RESULTS

Patients in the first group had a longer GC, i.e. their walking was slower. On the injured side, the differences from the control group were significant ($p<0.05$). In the second group, the differences were significant for both legs.

In the Flexion-extension movements at the hip joint on the injury side in the first group were reduced compared to the first-group healthy side and to the second-group injured side. The differences were statistically significant (p<0.05).

Knee joint movements were changed (Fig 1). In the first group, both flexion amplitudes were decreased compared to those of the healthy side and of the injured knee joint in the second group (p<0.05). Changes observed in the second group did not reach significance level, although the main flexion amplitudes on the injury side were also reduced compared to the healthy side. The main amplitude of flexion at swing phase was decreased on the injury side in the second group and increased on the healthy side in the first group (p<0.05).

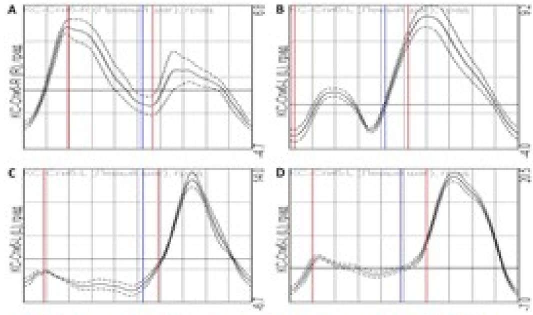

Figure 1. Consecutive stages of KJ function following ACL tear: A, B, C, D.

DISCUSSION

As shown by the study, not only knee function decreased in the first days post-trauma, but also the function of the lower limb as a whole. As a result, functional asymmetry developed. The kinematics of the involved joints changed in accordance with slower walking.

In the first days after trauma, the injured KJ developed a condition that can be described as “functional immobilization”. The movements in the joint while walking were of small amplitude, rocking, and occurred only under load. The amplitude of the main flexion in the swing phase was reduced (Fig. 1). Later on, flexion in the weight acceptance phase recovered. The last to respond was the extension amplitude in the midstance period. By the fourth week, the pattern of movements in the knee joint recovered to normal, although peak amplitudes remained reduced.

The obtained results suggest that the periods of post-trauma condition of the joint (acute, subacute, chronic) should be defined not only in terms of time (because individual response may vary significantly), but also taking account of the functional condition of the joint.The conducted study showed that acute ACL injury is associated with an overall reduction of function of the entire lower limb. The KJ at this stage is actively blocked by the strained muscles. The term “instability” of the KJ needs clarification.

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DISCLOSURE STATEMENT

The authors declare no conflict of interest.

Sagittal Plane Motion during Different Squat Tasks in Patients with Femoroacetabular Impingement

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INTRODUCTION

Femoroacetabular impingement (FAI) is a condition in which extra bone grows on either the acetabulum or the femoral head. This causes the bones to not fit properly together which can limit the range-of-motion (ROM) that can be achieved by the hip. Biomechanically, squatting is used to replicate daily tasks and can be used to test the overall ROM of the hip and knee joint.[1] The purpose of this study was to determine which squatting technique would elicit the greatest amount of hip and knee ROM and the greatest squat depth for biomechanical assessments.

CLINICAL SIGNIFICANCE

Participants may restrict the depth of their squat to maintain comfort and avoid pain. If squatting is used as a functional assessment or outcome measure, it is important to consider which squat technique elicits the largest amount of hip and knee motion.

METHODS

All participants were enrolled in an IRB approved research study and scheduled to undergo unilateral hip preservation surgery for FAI. Pre-operative kinematic data were collected while participants completed three squatting techniques. During the hold squat, participants were instructed to maintain the squat at their lowest possible position for three seconds. For the standard squat, each participant squatted to their lowest possible position and immediately returned to standing upright. For the target squat, a 15.5cm platform was placed behind the participant who was instructed to attempt to reach but not sit on the platform.[2] Peak sagittal plane kinematics of the trunk, pelvis, hip, knee, and ankle were determined during the squat. Maximum squat depth was calculated by dividing the displacement of the hip joint center (HJC) by the highest position of the HJC. Parametric statistics were used to evaluate comparisons between squatting techniques for Group 1. A One-way Repeated Measures ANOVA followed by a post-hoc paired t-test was used for comparison of Group 2 ($\alpha = 0.05$).

RESULTS

Forty-six participants that completed the hold and standard squats were included in Group 1 (16.5 $\pm$ 1.7 years, 35 females). Group 2, a subgroup of Group 1, consisted of twenty-five participants (16.5 $\pm$ 2.2 years, 20 females) who completed all three types of squats. The surgical breakdown of Group 2 (21 arthroscopic, 3 surgical hip dislocation, and 1 combined approach) was not as balanced as Group 1 (23 arthroscopic, 22 surgical hip dislocation, and 1 combined approach).

Overall, Group 1 sagittal plane motion of the hip, knee and ankle were significantly different between the two squatting techniques (Table 1). During the standard squat, maximum sagittal plane motion was greater than the hold squat for the hip (4.2°), knee (10.2°) and ankle (1.4°). While peak ankle dorsiflexion is statistically, due to the minimal difference we do not consider it clinically significant. Maximum squat depth was also greater during the standard squat with participants achieving 6.6% more HJC displacement. Comparing the hold and standard squat types, Group 2 showed similar results as Group 1 at the hip, knee, and ankle with all displaying increased peak values during the standard squat. (Table 2). While the standard squat had increased trunk tilt when compared to the hold squat, the greatest amount of trunk tilt was seen during the target squat in Group 2. The least amount of squat depth was achieved during the hold squat and the greatest during the target squat.

DISCUSSION

Participants may have restricted the depth of the hold squat to assure comfort and ability as they held the squatting position for three seconds. The amount of joint ROM and overall squat depth noticeably progressed with the deep squat, however, it was greatest in the target squat. Providing a target maximizes overall squat depth which may elicit different mechanics and compensations which may not be present during an untargeted hold or standard squat. Therefore, if biomechanical testing is used as a functional assessment or outcomes measure, a target squat maybe used to achieve the deepest squat and the greatest amount of hip flexion.

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DISCLOSURE STATEMENT

All authors have no conflicts of interests to disclose.

Table 1. Kinematic data of the affected limb of Group 1 during the squatting tasks.

AFFECTED LIMB	Group 1 mean (SD)		Hold vs Standard
	Hold	Standard	*p*-value
Max Trunk Tilt (°)	28.5 (12.0)	29.8 (13.2)	0.312
Max Pelvic Tilt (°)	2.7 (3.8)	2.3 (3.4)	0.062
Max Hip Flexion (°)	87.7 (14.8)	91.9 (14.1)	**0.005**
Max Knee Flexion (°)	90.2 (20.6)	100.4 (19.4)	**<0.001**
Max Ankle Dorsiflexion (°)	32.0 (6.0)	33.4 (5.9)	**0.001**
Max Squat Depth (%)	36.0 (13.0)	42.6 (12.8)	**<0.001**

Table 2. Kinematic data of the affected limb of Group 2 during the squatting tasks.

AFFECTED LIMB	Group 2 mean (SD)			One-way Repeated ANOVA	Hold vs Standard	Target vs Hold	Target vs Standard
	Hold	Standard	Target	*p*-value	*p*-value	*p*-value	*p*-value
Max Trunk Tilt (°)	27.9 (12.5)	31.5 (12.9)	39.2 (13.8)	**<0.001**	**0.002**	**<0.001**	**<0.001**
Max Pelvic Tilt (°)	2.6 (2.7)	2.2 (2.5)	2.8 (2.6)	0.323			
Max Hip Flexion (°)	87.2 (17.1)	94.5 (14.4)	100.9 (14.4)	**<0.001**	**<0.001**	**<0.001**	**<0.001**
Max Knee Flexion (°)	85.0 (18.4)	97.9 (19.7)	108.0 (21.2)	**<0.001**	**<0.001**	**<0.001**	**<0.001**
Max Ankle Dorsiflexion (°)	30.4 (5.6)	32.0 (5.5)	29.8 (5.3)	**0.003**	**0.016**	0.403	**0.001**
Max Squat Depth (%)	34.0 (12.0)	42.7 (12.9)	50.4 (14.2)	**<0.001**	**<0.001**	**<0.001**	**<0.001**

Similar Biomechanics During Change of Direction in Adolescents with Contact Versus Non-Contact Anterior Cruciate Ligament Injury

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Introduction

Patients who sustain non-contact anterior cruciate ligament (ACL) injuries are thought to be predisposed to injury due to deficient biomechanics or neuromuscular control[1]. In contrast, patients who sustain contact ACL injuries may have been injured due to unlucky trauma and may not have poor biomechanics. The purpose of this study was to compare biomechanics during change of direction movements between patients who injured their ACL through a contact mechanism and those who were injured with a non-contact injury mechanism. We hypothesized that patients who sustained an ACL tear with a contact injury mechanism would demonstrate better biomechanics (greater shock absorption and less dynamic limb valgus) than patients with a non-contact injury mechanism.

Clinical Significance

The results of this study suggest that all patients post ACL reconstruction have potentially modifiable biomechanical risk factors for re-injury regardless of injury mechanism. Motion analysis can be used to evaluate movement patterns prior to return to sport so any deficiencies in biomechanics or neuromuscular control can be identified and addressed.

Methods

15 adolescents (age 10-18 years) with contact ACL injury (4 female; mean age 15.5, standard deviation (SD) 2.1 years) and 94 with non-contact ACL injury (11 female; mean age 15.6, SD 1.9 years) underwent biomechanical assessment in our sports and motion analysis laboratory 6-12 months (mean 7.5, SD 1.3) after ACL reconstruction. Subjects performed forward-backwards (deceleration) and lateral (side shuffle) change of direction tasks while 3D motion analysis data were collected using a 10-camera Vicon motion capture system and AMTI force plates. A 6-degree of freedom marker set was used, and modeling was performed in Visual3D. Kinematic and kinetic variables reflecting dynamic limb valgus (frontal and transverse plane) and shock absorption (sagittal plane) were compared between patients who had contact vs. non-contact injury mechanisms using 2-tailed t-tests.

Results

No significant differences were observed in any of the kinematic or kinetic measures between the contact and non-contact groups (Table 1).

Table 1: Comparison of kinematics and kinetics between contact and non-contact ACL injury groups

	Deceleration			**Lateral Shuffle**		
	Non-Contact	Contact	P-value	Non-Contact	Contact	P-value
SHOCK ABSORPTION						
Max hip flexion	75.3 (15.2)	76.9 (16.4)	0.72	68.4 (14.6)	71.9 (13.6)	0.39
Max knee flexion	65.2 (14.1)	68.8 (20.9)	0.39	61.4 (13.1)	65.2 (13.5)	0.31
Max ankle dorsiflexion	-5.5 (7.1)	-2.3 (2.2)	0.12	16.0 (7.5)	18.2 (8.9)	0.32
Max hip flexion moment	2.8 (1.5)	2.5 (0.9)	0.58	2.07 (0.52)	2.10 (0.63)	0.89
Max knee flexion moment	1.3 (0.5)	1.2 (0.7)	0.55	1.20 (0.50)	1.25 (0.45)	0.72
Max ankle dorsiflexion moment	0.84 (0.22)	0.82 (0.29)	0.80	1.07 (0.30)	1.14 (0.55)	0.52
Energy absorption at hip	0.66 (0.43)	0.56 (0.39)	0.39	0.50 (0.26)	0.45 (0.25)	0.50
Energy absorption at knee	0.50 (0.35)	0.44 (0.38)	0.52	0.38 (0.26)	0.38 (0.19)	0.96
Energy absorption at ankle	0.17 (0.11)	0.14 (0.06)	0.25	0.41 (0.19)	0.42 (0.25)	0.84
DYNAMIC LIMB VALGUS						
Max hip internal rotation	7.8 (7.4)	5.1 (7.8)	0.19	13.7 (8.8)	9.3 (7.5)	0.07
Max hip adduction	1.9 (6.4)	2.6 (4.5)	0.70	-18.3 (8.3)	-17.2 (7.0)	0.65
Min knee varus	-1.1 (4.8)	-3.8 (6.6)	0.05	-2.4 (5.1)	-3.5 (5.4)	0.44
Min knee varus moment	-0.34 (0.34)	-0.32 (0.32)	0.82	-0.85 (0.69)	-0.81 (0.93)	0.84

External moments are reported. Angles are expressed in degrees, moments in N/kg, energy absorption in J/kg.

Discussion

Contrary to expectations, in patients recovering from ACL reconstruction, the contact injury group did not have better biomechanics than the non-contact injury group. This may be due to both groups engaging in similar physical therapy and rehabilitation programs, focusing on improving their strength and movement patterns. It may also be due to the contact injury group having similar pre-injury biomechanics to the non-contact group but happening to be injured by a contact mechanism. These results suggest that all patients post ACL reconstruction have potentially modifiable risk factors for re-injury and should have their biomechanics evaluated so any deficiencies in movement patterns, biomechanics, and neuromuscular control can be rectified prior to return to sport regardless of injury mechanism.

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Cognitive and Functional Performance in Adolescents Following Concussion: From Clinical Presentation to Initiation of Return to Play

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INTRODUCTION

The prevalence of concussions has increased 60% in the last decade, with adolescents as the primary contributor [1]. The adolescent population is especially important to investigate due to the elevated risk for more severe symptoms and prolonged recovery [2]. The effect of concussion on balance is commonly examined in clinic using observation-based screens, such as the Balance Error Scoring System (BESS), to assess severity of concussion-related functional deficits [3]. However, more objective measures, such as center-of-pressure (COP), have shown that individuals can have balance and postural stability deficits 1-month post-concussion, despite having BESS scores similar to non-concussed controls [4]. Additionally, dual-task gait evaluations have also been used to assess the effects of concussion on cognitive and motor function [5]. Although these reports have shown inconsistent findings, dual-task gait evaluations remain highly consistent across testing sessions with demands affected by task complexity [6].

CLINICAL SIGNIFICANCE

Determine whether cognitive deficits and functional performance variations in adolescents who have sustained a concussion are evident at the time of clinical clearance to initiate the return to play (RTP) protocol.

METHODS

Patients who were diagnosed with a concussion in the sports medicine clinic and met inclusion criteria were invited to participate in this IRB approved study. Patients enrolled were seen for testing within 10 days of their clinical diagnosis (V0) and again once cleared by their physician to begin the RTP protocol (V1). Testing included a balance assessment, simple and dual-task gait, as well as fine motor reaction time. The balance assessment included performing the BESS while standing on a force platform for 20 seconds with eyes closed during 6 conditions: double-leg, single-leg, and tandem stance on both a firm and foam surface. COP max sway in the medial/lateral (ML) and anterior/posterior (AP) directions were analyzed for the first bout of 5 seconds in which the subject remained in the testing position. The number of errors were documented for each condition according to the BESS score card and totaled for each surface [3].

For the simple gait task, subjects were instructed to walk at a self-selected speed for 10m. Dual-task gait consisted of walking 10m while reciting 3-digit numbers in reverse order. Temporal parameters, including cadence, walking speed, stride length, step width, single limb support and double limb support were calculated using bilateral heel and toe markers. Performance was recorded for dual-task gait as the percentage of 3-digit numbers recited correctly. Lastly, fine motor reaction time was measured using a custom module that measured the time between a visual cue and when the subject pushed a button. The average time of 10 trials was computed for each visit. A paired student's t-test was used to compare variables across visits (α=0.05).

RESULTS

Twenty-nine subjects were seen at both visits, results are shown in Tables 1 and 2. On the firm surface, ML max sway during double-leg stance decreased while there was no reduction in AP max sway from V0 to V1. There were no significant changes in max sway for double-leg stance on the foam surface across visits. Sway measures for tandem stance on both surfaces were not significant and an insufficient number of participants were able to complete single-leg stance at both visits to analyze. The number of errors was reduced at V1 on the foam but not for the firm surface. There was no change in cadence during either simple or dual-task gait, however, walking speed did increase with both gait tasks due to a significant increase in stride length. Performance during dual-task gait did not significantly improve (V0 78%, V1 81%; p=0.378) and there were no changes in fine motor reaction time (V0 460.4ms, V1 434.9ms; p=0.219).

Balance		Visit 0	Visit 1	*p*
BESS Firm	# Errors	2.9 (2.3)	2.2 (1.9)	0.059
	DS ML Max Sway (mm)	13.8 (5.5)	10.9 (9.3)	**0.022**
	DS AP Max Sway (mm)	12.1 (6.3)	12.4 (7.1)	0.796
	TS ML Max Sway (mm)	25.3 (11.6)	23.3 (13.0)	0.378
	TS AP Max Sway (mm)	27.7 (20.4)	25.5 (17.4)	0.585
BESS Foam	# Errors	7.0 (1.8)	6.1 (2.0)	**0.015**
	DS ML Max Sway (mm)	24.6 (7.8)	26 (9.4)	0.436
	DS AP Max Sway (mm)	30.7 (9.9)	28.8 (13.8)	0.448
	TS ML Max Sway (mm)	46.8 (22.7)	39.7 (31.8)	0.305
	TS AP Max Sway (mm)	43.9 (16.4)	38.8 (31.8)	0.473

Table 1: Balance measures at V0 and V1 (mean (SD))

	Simple Gait			Simple Gait Dual-Task		
	Visit 0	Visit 1	*p*	Visit 0	Visit 1	*p*
Cadence (steps/min)	115.99 (10.19)	117.5 (10.71)	0.261	112.22 (10.79)	114.15 (9.99)	0.105
Walking Speed (m/s)	1.22 (0.18)	1.27 (0.15)	**0.007**	1.11 (0.20)	1.21 (0.16)	**<0.001**
Stride Length (m)	1.26 (0.12)	1.3 (0.11)	**0.001**	1.18 (0.14)	1.27 (0.11)	**<0.001**
Step Width (m)	0.119 (0.03)	0.122 (0.03)	0.405	0.126 (0.03)	0.121 (0.03)	0.301
Single Limb Support Left (%GC)	41.97 (1.23)	42.53 (1.25)	0.064	40.63 (2.04)	41.15 (2.27)	0.297
Single Limb Support Right (%GC)	41.87 (1.61)	42.69 (1.77)	**0.038**	40.49 (2.72)	41.35 (1.94)	0.073
Double Limb Support Left (%GC)	15.92 (2.74)	14.96 (2.57)	0.062	18.74 (3.90)	16.57 (3.87)	**0.009**
Double Limb Support Right (%GC)	15.41 (2.64)	14.87 (2.89)	0.345	18.66 (3.47)	16.51 (3.10)	**0.002**

Table 2: Temporal measures at V0 and V1 for Simple and Dual-task Gait (mean (SD))

DISCUSSION

Improvements were seen in double-leg stance ML max sway on the firm surface when subjects were cleared to begin the RTP protocol compared to the initial visit. Additionally, participants increased their stride length and subsequently walking speed during simple and dual-task gait conditions. The total number of errors on the foam was reduced at V1, but performance during dual-task gait or fine motor reaction time did not change. Further work is needed to compare these findings to an age-matched control cohort.

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PERFORMANCE MEASURES ASSOCIATED WITH SPORTS-SPECIALIZATION

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INTRODUCTION

Sports specialization is becoming more prevalent in youth sports, which has led to reduced cross-training, the elimination of an off-season, and consequently, more overuse injuries [1]. While difficult to accurately assess, literature suggests that overuse injuries account for 46-50% of all athletic injuries [2]. Numerous reports have also shown common changes in biomechanics and performance post-injury, however, it is unclear exactly how sports-specialized athletes move and perform differently than non-specialized athletes pre-injury. Therefore, the purpose of this study was to complete a preliminary analysis evaluating the relationship between performance, age, and number of years specialized.

CLINICAL SIGNIFICANCE

To investigate which measures of strength, balance, speed, and stability are influenced by sports-specialization in two different age groups.

METHODS

Thirty-five female athletes were recruited to participate in the SAFE (Specialized Athletes Functional Evaluation) Program. All athletes recruited play at an elite level and were enrolled in an IRB approved research study. Knee strength, dynamic balance, sprint speed, and single-leg hop (SLH) distances were collected, along with age, height, weight, and number of years specialized in soccer. Knee flexion and extension strength was measured using a Biodex System 3 collected at 120° per second, and peak torque across five repetitions was recorded. Dynamic balance was assessed using the Lower Quarter Y-Balance Test (YBT), which measures the reach distance of the non-stance limb in three directions and normalizes each distance by leg length. Lastly, sprint speed was recorded for 10m and 20m distances. Kinematic data were also captured for a variety of dynamic tasks, such as the SLH.

Data was divided into two sub-groups based on age (Group 1: 9-13 yrs., Group 2: 15-18 yrs.). Mean and standard deviations were calculated for all variables, and unpaired t-tests were used for comparison between the two age groups (Table 1, $\alpha = 0.05$). Within each group, Pearson correlations were computed to assess which performance measures were significantly associated with age and/or number of years specialized (Table 2, $\alpha = 0.05$).

RESULTS

No significant differences were found between limbs for strength, balance, speed, or hop distance, therefore, only values from the dominant side were reported (all athletes were right-leg dominant). Group 2 performed significantly better than Group 1 in all performance tasks except for the YBT and 10m sprint. Additionally, the number of years each group has been specialized in soccer was not significantly different.

Age and number of years specialized were not significantly correlated for Group 1 ($R^2 = -.31$, $p=0.18$) or Group 2 ($R^2= 0.20$, $p=0.47$). For Group 1, all variables were significantly associated with age, except for balance scores and sprint speeds. However, sprint speeds were able to distinguish between number of years specialized. Furthermore, although balance scores were not different between the two groups, they were significantly associated with age in Group 2, along with strength and 20m sprint speed. SLH distance and 20m sprint speed were the only two measures significantly associated with years specialized in Group 2.

Table 1: Demographic and performance data for both age groups (mean ± SD).

Variable	**Group 1 (n = 15)**	**Group 2 (n = 20)**	***p*-value**
Age (yrs)	11.31 ± 1.59	16.75 ± 1.06	**< 0.001**
Specialized (yrs)	6.47 ± 2.50	6.95 ± 3.59	0.661
Height (m)	1.50 ± 0.13	1.67 ± 0.08	**< 0.001**
Weight (kg)	41.87 ± 10.94	64.59 ± 15.79	**< 0.001**
Flexion (Nm)	36.17 ± 12.07	52.56 ± 12.78	**< 0.001**
Extension (Nm)	68.65 ± 19.31	91.68 ± 22.21	**0.003**
YBT Score	100.27 ± 8.46	100.75 ± 10.32	0.884
10m Sprint (s)	2.59 ± 0.35	2.41 ± 0.23	0.076
20m Sprint (s)	4.43 ± 0.73	3.99 ± 0.26	**0.018**
SLH Dist. (m)	0.97 ± 0.19	1.19 ± 0.22	**0.004**

Table 2: Correlations (R^2) with age and years specialized for each group (p<0.05 in bold).

	Group 1 (n = 15)		**Group 2 (n = 20)**	
Variable	Age	Yrs. Specialized	Age	Yrs. Specialized
Height (m)	**0.77**	0.41	0.06	-0.30
Weight (kg)	**0.54**	0.47	0.25	0.04
Flexion (Nm)	**0.63**	0.22	**0.62**	-0.37
Extension (Nm)	**0.64**	0.48	**0.46**	-0.42
YBT Score	-0.39	-0.34	**0.82**	-0.29
10m Sprint (s)	-0.43	**-0.66**	-0.42	0.33
20m Sprint (s)	-0.50	**-0.63**	**-0.61**	**0.54**
SLH Dist. (m)	**0.53**	-0.30	0.36	**-0.47**

DISCUSSION

These results suggest that while certain performance measures are naturally dependent on age, specialization in a single sport, in this case soccer, may also impact performance. Furthermore, these trends may change as the athlete grows older. For example, only age significantly influenced SLH distance in the younger group while the same measure was influenced by years specialized instead of age in the older group.

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Energy Flow Analysis of Professional and Collegiate Baseball Pitchers

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INTRODUCTION

Elbow and shoulder injuries often occur in baseball pitchers due to overuse and the repetitive throwing motion, creating microtrauma in the musculoskeletal system that is unable to fully recover.[1] During the pitching cycle (PC), power is generated from the force of the back leg and up through the body in a sequential segmental energy transfer that follows a proximal-to-distal flow through the throwing arm.[2] Although there is an agreement amongst previous research these specific joint injuries are related to excessive surrounding forces[3], there has yet to be a consensus regarding what movements during the pitching motion have direct effect on these forces. The purpose of this study was to conduct a segmental power analysis of pitchers at both the professional and collegiate levels and compare the forces generated.

CLINICAL SIGNIFICANCE

Monitoring how energy flows throughout the PC 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 information may then be translated to pitching coaches for injury prevention.

METHODS

13 collegiate (age: 21.2 ± 2.3 years, height: 184.94 ± 8.11 cm, weight: 90.9 ± 13.5 kg,) and 13 professional pitchers (age: 22.8 ± 1.8 years, height: 185.87 ± 2.45 cm, weight: 92.4 ± 3.2 kg,) participated in the study. Inclusion criteria required subjects to be over 18 and no injury within the previous 12 months. Each subject completed informed consent and provided medical historyn prior to testing. Forty-seven reflective markers (12.5 mm diameter) were attached to the subjects at specific bony prominences. A system of 8 Raptor-E cameras (Motion Analysis Corporation, Santa Rosa, CA) was used to capture the motion of pitchers at 300 frames per second. The motion analysis system was set up surrounding a pitching mound, where each participant threw 10 fastballs. Velocity was recorded using a Stalker Sport 2 radar gun (Stalker Sports Radar, Richardson, TX), along with the location of each pitch. All data was exported and processed with Visual 3D software (C-Motion, Germantown, MD) using a previously developed biomechanical pitching model.[4] To minimize influences from height and weight differences, the torques were normalized by weight and height [Normalized = Torque (Nm) / (BW(N)*Height(m))]. This created a unitless numeric value that could be compared across groups. Statistical analysis was completed using SPSS with a significance level of .05 used.

RESULTS

Independent t-tests found ball speed to be significantly different ($p<.001$) between groups (Table1), and that professionals had significantly higher upper arm power at both maximum external rotation (MER) ($p=.006$) and ball release (BR) ($p=.047$). A multiple stepwise linear regression completed on all pitchers from both groups showed that ball speed was influenced by upper arm power (p=.02), and that absolute maximum elbow varus torque (EVT) was influenced by percent timing of torso flexion (p=.005) and percent timing of EVT (p=.049). A multiple stepwise linear regression on the professional group found absolute EVT to be influenced by percent time upper arm power (<.001) and torso power (p=.027). Shoulder internal rotation torque (SIRT) was found to be influenced by upper arm power (p=.018) and percent timing of upper arm power (p<.001). A multiple stepwise linear regression on the collegiate group found ball speed to be influenced by percent timing torso flexion(p=.001), absolute EVT was influenced by percent timing torso rotation (p=.019), and SIRT was influenced by percent timing torso rotation (p=.029). There were no statistical differences between groups for age, height or weight.

Table 1: Key data points compared across the two groups via independent t tests

	Collegiate (n=13)	Professional (n=13)	P
Ball Speed, m/s	34.64±1.87	38.63±.96	<.001
Absolute EVT, N·m	124.20±29.07	118.39±19.03	.553
Normalized EVT	.0757±.0148	.0703±.01085	.302
Absolute SIRT, N·m	116.79±25.01	115.84±20.39	.916
Normalized SIRT	.0711±.011	.0688±.0117	.614
Torso Power, W/kg	18.20±7.12	23.45±12.77	.807
Upper Arm Power, W/kg	16.78±12.16	24.64±16.00	.172
Wrist Power, W/kg	31.06±12.36	27.95±8.33	.460

Data are shown as mean ± SD
Normalized variations = Absolute value/ [Height (m) x Body Weight (kg)]

DISCUSSION

The regression analysis suggests the significant factor influencing ball speed is maximum upper arm power which, although insignificant, was found more commonly at greater levels amongst professional pitchers. In support of this relationship, data showed professionals threw at a significantly greater ball speed than collegiate players. It is notable both groups were still found to have relatively similar EVT and SIRT joint stresses. Though the two groups also reached similar levels of maximum segmental powers, the moment during the pitching cycle at which they reach these peak powers may differ. Proper timing of specific pitching mechanics may be a key difference between professional and collegiate levels, resulting in the lower joint torques and greater ball speed. Future research will investigate power and timing throughout the entire pitching cycle to more specifically determine the variables influencing ball speed and joint torques.

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DISCLOSURE STATEMENT

The authors have nothing to disclose.

JOINT KINETICS IN UNDERSTANDING RUNNING PATHOLOGY: A CASE STUDY FOR AN ADOLESCENT WITH PAIN

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PATIENT HISTORY

Patient is a 17-year-old male, who had a transverse fracture of his tibia and fibula two years prior at 15 years of age when playing baseball. Surgical fixation took place 4 days following with plate and screws on both the tibia and fibula followed by casting and crutch walking for two weeks followed by boot for two months with full healing. He had 8 months of PT following surgery before approval to return to baseball. Due to pain while loading in running he was unable to perform at his pre-injury level. Three months later he was seen for a comprehensive gait analysis to gain insight into the possible mechanisms for the ongoing pain during running. Parent and patient goal was pain resolution and to attain preinjury running speed and agility.

CLINICAL DATA

A comprehensive gait analysis was completed both walking and running. Selected clinical exam findings are summarized in Table 1. Pain was located 1/3 the way down the anterior portion of the right shank and was graded 3-4/10 when at its worst. Additional assessments revealed difficulty with single leg hopping and squatting (eccentric loading of the quadriceps) on the right side only.

Table 1: Selected clinical exam and temporal findings for pre and post-surgery.

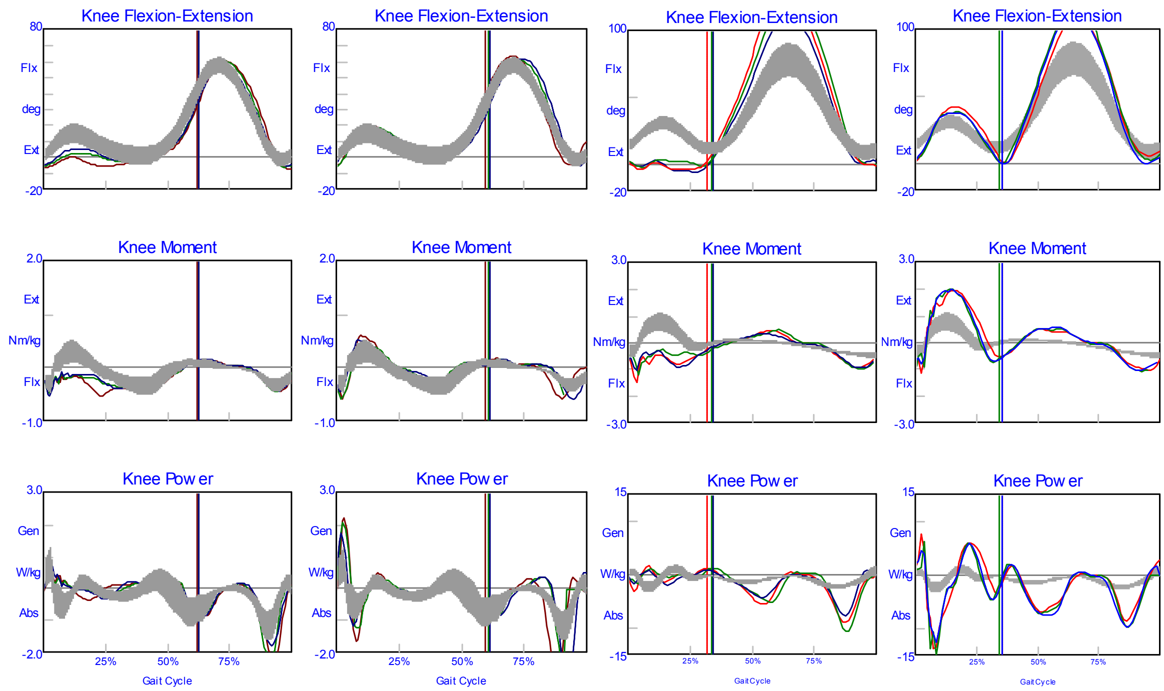
	Right	Left
Hip flexion strength	5/5	5/5
Knee flexion strength	5/5	5/5
Knee extension strength	5/5	5/5
Ankle plantar flexion strength	5/5	5/5
Knee extension (deg)	0	0
Popliteal angle (deg)	-55	-45
Ankle dorsiflexion knee 0deg flexion (deg)	5	5
Ankle plantar flexion strength	5/5	5/5
Ankle dorsiflexion strength	5/5	5/5
Quads Girth (20 cm prox to patella) (cm)	57.8	60.0
Toe off walking range (% gait cycle)	62.2-62.5	59.3-61.7
Toe off running range (% gait cycle)	33.7-34.1	34.1-35.7

MOTION DATA (C94250)

During level shod walking, motion analysis data showed a reduction in knee flexion loading and associated quadriceps avoidance moment pattern on the right in comparison to the left side (Fig. 1). When running, this asymmetry was more remarkable with no right knee flexion in loading with a continued quadriceps avoidance pattern during loading of the right side only (Fig. 1). Kinematic and kinetic data was very consistent stride to stride. The right side ankle showed no compensatory increased absorption during loading (not shown).

TREATMENT DECISION AND INDICATIONS

Physical therapy to work on eccentric and concentric strengthening of the right quadriceps. Indications: right knee reduction in loading response flexion in comparison to the left and associated quadriceps avoidance during both walking and more significantly in running. Ankle kinematics and kinetics were symmetric during running (not shown) indicating that he did not compensate for reduced knee loading at the ankle on the right side. Goal: to improve eccentric loading capacity of the quadriceps muscle and distribute loading impact through both the knee and ankle thus reducing the load on the right tibia.



Right Left Right Left

Walking Running

Figure 1: Comparison of the right and left side (three gait cycles each) sagittal plane knee kinematics and kinetics during walking and running.

TREATMENT OUTCOMES

The patient is currently undergoing rehabilitation to address the asymmetry in knee eccentric strength to allow muscle based absorption of landing loads during running on the right side.

SUMMARY

Comprehensive gait analysis allows objective measurement of walking and running function that is not possible in the clinic setting. The patient showed asymmetry in knee sagittal plane kinematics and kinetics both in walking and more so in running. The decreased eccentric loading at the right knee was not absorbed by increased right ankle absorption. These gait findings led to assessment of eccentric strength immediately following the gait analysis testing. The isometric strength symmetry did not provide evidence of this isokinetic strength asymmetry. The gait analysis results, specifically knee sagittal plane kinetics, helped to identify ongoing impairment that required rehabilitation.

DISCLOSURE STATEMENT

The authors have no conflicts of interest to disclose.

Biomechanical Performance and Limb Asymmetry Among Youth Athletes Recovering from Anterior Cruciate Ligament Reconstruction.

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Introduction: The incidence of anterior cruciate ligament (ACL) injuries among adolescent athletes is increasing, and reinjury rates after an ACL reconstruction (ACLR) are relatively high[1]. Biomechanical and motion capture analysis has been used after ACLR in order to assess readiness to return to sports[1,2]. In the current literature, landing tasks, such as the single leg drop jump, have been used to understand hip, knee, and ankle movement during these tasks[1,2]. While there is an abundant amount of literature, more research is required to fully understand changes in biomechanics and movement quality during these functional tasks after an ACLR surgery[1,2]. The purpose of this study was to analyze biomechanics of functional task after an ACLR surgery.

Clinical Significance: This study provides further insight to the specific tasks that should be analyzed, and the level of performance necessary to safely to return to sports after ACLR.

Methods: Patients underwent ACLR surgery, and a standard Physical Therapy rehabilitation program. As a part of the clinical evaluation, each completed a series of functional tasks. Demographic information, surgical procedure(s), concomitant injury, and time from injury to surgery were collected. Subjects who completed the assessment greater than 9 months post-surgery and before returning to sport were included in this analysis were included in this analysis. Subjects were excluded based on previous surgery to the involved or uninvolved leg or if their assessment was before 9 months. The biomechanical assessment was performed using a 13-camera Vicon motion capture system and Nexus software, in addition to custom analysis software written in Visual 3D and LabVIEW. During the analysis, 40 markers were placed on the whole body. Patients were asked to perform multiple tasks associated with the standard of care protocol after an ACLR surgery at our institution, including a single leg squat and a single leg drop jump task. Once all data was collected, means and SD were used to analyze categorical variables. Additionally, a comparison between surgical and non-surgical limb performance on each task was performed using paired samples t-tests.

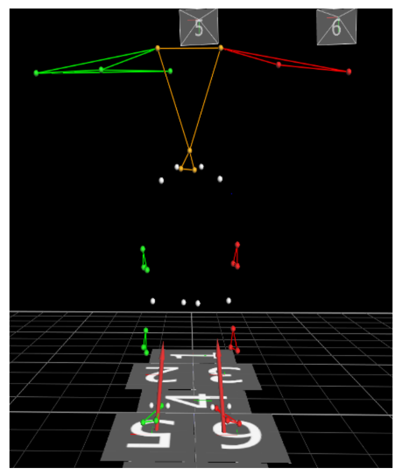

Figure 1: Marker Placement for Motion Analysis

Results: A total of 19 subjects (mean age=16.4±1.7 years, 45% were female) completed the assessment. Roughly 60% had right limb surgery, and 85% underwent primary quadriceps tendon–patellar bone autograft (5% hamstring autograft; 10% contralateral quadriceps tendon–patellar bone autograft). Data was available for 13 patients on the single leg drop jump test, and for 6 patients on the single-leg squat test. At maximum knee flexion during the single leg drop jump test, participants demonstrated significantly greater trunk tilt on the surgical side compared to the non-surgical side. All other variables were not statistically significant (Table 1).

Discussion: The difference in trunk tilt during the single leg drop jump test between surgical and non-surgical leg may relate to the patient's use of compensatory mechanisms. For a patient to achieve the same amount of knee flexion on both the surgical and non-surgical leg, the surgical leg must compensate, as they have not fully healed from the ACLR surgery and their muscle strength limits their range of motion. This manifests as increased trunk tilt on the surgical side at maximum knee flexion. While these results support past research, our results are still preliminary. Further research is required to determine the significance of differences in trunk tilt during a single-leg drop jump test.

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Table 1: Biomechanical Analysis

Variable	Surgical side	Non-surgical side	P value
Single-leg squat test (n=6)			
Maximum knee flexion (degrees)	91.5 (23.2)	95.0 (22.2)	0.53
Trunk tilt at maximum knee flexion (degrees)	46.9 (11.7)	43.1 (9.8)	0.10
Trunk obliquity at maximum knee flexion (degrees)	-6.2 (2.7)	-6.7 (4.7)	0.82
Trunk rotation at maximum knee flexion (degrees)	0.6 (4.7)	-0.5 (3.1)	0.60
Single-leg drop jump test (n=13)			
Maximum knee flexion (degrees)	68.8 (11.8)	70.2 (10.3)	0.46
Trunk tilt at maximum knee flexion (degrees)	41.6 (13.6)	38.4 (12.6)	0.04*
Trunk obliquity at maximum knee flexion (degrees)	-5.7 (5.6)	-6.5 (5.2)	0.70
Trunk rotation at maximum knee flexion (degrees)	-3.5 (6.3)	0.1 (5.9)	0.09

*=Significance ($p<0.05$)

HIP STRENGTH INFLUENCES GROUND REACTION FORCE ATTENUATION ON A SIDE LEAP IN COLLEGIATE DANCERS

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INTRODUCTION

Vertical ground reaction forces were found to be considerably higher during the landing phase (compared to take off) during common dance maneuvers. In 3D kinematic analysis, Kalichova et al. reported a large range of upper extremity movement outside of the center of gravity on the landing phase of a grand jete leap, essentially causing the landing phase to be unstable and implying a lack of lower extremity strength and control [1]. In a musculoskeletal model, Lewis et al. noted that progressively weakening the gluteal muscle group by decreasing maximum force production capabilities resulted in increased superior and anterior hip joint force during hip extension [2]. Compensation appeared to be provided by increased muscle activity in the semimembranosis, gluteus medius, vastii muscles and tensor fascia latae, which may relate to injury [2]. Valente et al. similarly modeled sensitivity of joint contact force components to perturbation of the force generating capacity of ipsilateral hip abductor muscles during a walking gait and found that reductions, particularly in the anterior compartment of the gluteus medius, increased risk of overloading at both the hip and knee joints [3]. Although static Q-angle has been found to be a poor independent predictor of lower extremity injury risk, Nguyen et al. found that greater tibio-femoral angle and femoral anteversion were significant predictors of greater Q-angle in both males and females [4], suggesting interplay between Q-angle and hip and shank positioning. Collegiate female soccer players with a supple planus foot type foot type showed decreased hip muscle activation associated with increased vertical ground reaction force during landing [5]. and females with supple planus feet showed 49% lower biceps femoris and gluteus maximus activation and 31% greater medial shear ground reaction forces post fatigue, compared to those with rectus feet [6].

CLINICAL SIGNIFICANCE

The vast majority of dance-related hip injuries are due to overuse, muscular compensation secondary to strength imbalances, and lower extremity misalignment. Altered landing mechanics may decrease force attenuation capacity at ground contact. Evaluation of jump landing strategies exhibited by dancers and differences elicited post fatigue on a side leap maneuver might elucidate the relative contribution of these factors to injury risk. The purpose of this study was to examine pre to post fatigue ground reaction force attenuation differences potentially influenced by strength and alignment factors along the lower extremity kinetic chain.

METHODS

16 healthy experienced female dancers from a university dance team participated in a cross-sectional design. Independent variables were strength and agonist-antagonist strength ratios for hip extensors, flexors, abductors, adductors, lateral and medial rotators, and knee

extensors, q-angle, foot type and time. Dependent variables were peak vertical force, rate of loading, and anterolateral shear force composite.

RESULTS

Independent t-test showed dancers with higher composite hip strength scores had significantly lower peak normalized vertical force (p =.01, t = 2.16) and vertical rate of loading (p =.004, t = 2.16) pre-fatigue on a side leap landing. No other group differences in strength, static Q-angle, foot mobility or fatigue were statistically significant.

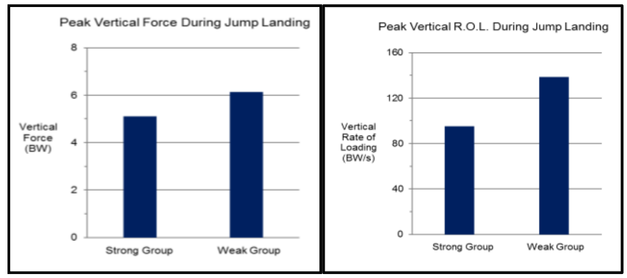

Figure 1. Ground reaction force attenuation difference by strength

DISCUSSION

Hip-strong dancers were better able to attenuate vertical force at ground contact pre-fatigue. Traditional analyses on dance-related impact landings have examined vertical components and associated alignment flaws. However, lateral type landings a side leap might redirect some of the landing force attenuation load from sagittal and vertical components to lateral shear force. Future research models should consider multi-directional forces imposed at ground contact during complex landing maneuvers.

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DISCLOSURE STATEMENT

Authors have no conflicts of interest to disclose.

EFFECT OF TREADMILL-BASED RESISTANCE ON LANDING STRATEGY AND FORCE ATTENUATION IN FEMALE COLLEGIATE LACROSSE PLAYERS

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INTRODUCTION

ACL injuries remain prevalent, especially in team sports involving frequent cutting, jumping, landing, and pivoting movements throughout game or practice [1]. Sport comparison research has examined sex differences in ACL injury incidence, noting females to be highly at risk in basketball and soccer, with lacrosse trailing right behind [1].

A non-contact ACL injury typically occurs at foot strike when an athlete is decelerating, pivoting, landing, or responding to an unanticipated perturbation [2]. Femoral motion and position affect landing force attenuation and have been shown to be modifiable through strength training proximal to the knee. Targeting hip extensors acting eccentrically to attenuate ground force during landing can be logically linked to regulation of leg spring stiffness along the kinetic chain during weight-bearing activities.

Hip muscle activation can affect knee loading as upper body loads are transferred through the hip to the lower extremity [3]. Accordingly, endurance of hip muscles may play a vital role in neuromuscular control at the knee to maintain optimal sagittal plane alignment. Lower extremity muscle fatigue has been found to alter force production [4], proprioception [5], coordination [6], and landing kinematics [7], which might increase injury risk.
An abrupt braking strategy during landing, associated with greater hip and knee extensions due to hip extensor weakness and inability to effectively regulate the load at ground contact, limits force attenuation capacity of the quadriceps and hamstrings that normally occurs with knee flexion [4] and potentially subjects the joint to buckling [8].

CLINICAL SIGNIFICANCE

Novel methods to target and condition hip extensors have been proposed, with closed kinetic chain resistance exercises widely considered most functional in achieving adaptations relevant to knee stability and coordinated landing strategies. However, the use of treadmill-based resistance training for developing hip-specific strength in this context has not been investigated. Our purpose was to examine the effect of six weeks of modified incline treadmill-based resistance training on functional landing strategies, vertical ground force attenuation and knee and trunk flexion angles in female athletes, compared to active controls.

METHODS

15 healthy female intercollegiate lacrosse players (age = 19.5 ± 1.7 years, height = 1.65 ± .23m, weight = 59.33 ± 5.4kg) participated in a repeated measures, cross-sectional design and provided written informed consent. Independent variable was time (pre- and post- training). Dependent variables were reactive strength index (RSI), vertical ground force rate of loading (ROL), and knee and trunk flexion angles during drop jumps from a 30cm box. Training occurred on two non-consecutive days per week, over six weeks with treadmill set at 15

percent grade and progressive cable resistance load set initially at 40% of hip extensor strength average for a duration of 7 minutes per session.

RESULTS

Paired Samples t-tests showed a significant ($p=.007$) increase in RSI post training, specifically reflecting a 12.5% increase in RSI scores. No other group differences in ROL or knee or trunk flexion angles were statistically significant pre-to post training.

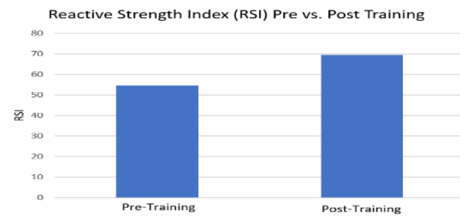

Figure 1: RSI Changes Pre-to Post Training.

DISCUSSION

Calculated as vertical jump height divided by pre-jump ground contact time, RSI differences reflect training adaptations to enhance transfer of eccentric landing force load to a propulsive outcome, demonstrating improved power and plyometric performance. Fluid transfer of landing force into a propulsive outcome at ground contact is pivotal to successful transitioning in sports, such as lacrosse, involving cutting, jumping, and pivoting movements. Future research should investigate the effect of strength, muscle activation and leg spring stiffness changes on dynamic knee restraint mechanisms protective against ACL injury.

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DISCLOSURE STATEMENT

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Medial-lateral hand location effect on motor control and neuromuscular activation during push-up exercise

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Introduction

The push-up is a popular exercise frequently used in several strength, stability, and conditioning training to develop neuromuscular endurance or stability modification. Although, despite familiarity with the push-up amongst different athletics, there is a lack of knowledge representing neuromuscular activation during push-up exercise among athletics, coaches, clinical specialists, and sports trainers [1]. The push-up can be executed with many different hand locations [2]. These variants may arise from moment arm lengths [3]. The aim of the current study is to evaluate neuromuscular activation of the selective upper limb muscles during five medial-lateral hand location push-up exercises.

Clinical Significance

Weakness in the shoulder complex causes insufficient stabilization of the scapula, changing its motion and basically locating the particular for future injury risk. The push-up exercise can increase muscle activation of some upper limb muscles and these can develop stabilization of scapular in many sport-specific motions, including badminton, tennis serves, softball pitching, and golf swings [4]. In addition, the push-up is widely recommended for the recruitment of multiple shoulder complex and scapular muscles, especially in rehabilitation and sports medicine [5].

Methods

Fifteen professional fitness athletics males (29.1 ± 6 years; 173 ± 7 m; 68 ± 9.23 kg) participated in the current experiment. Athletics were free from any mental or physical problems. Ten high-speed cameras (Vicon, Oxford, UK) were used to collect three-dimensional motion of subject at a sampling rate of 100 to describing push-up motions. Surface EMG was recorded at 2000 Hz (Myon Ltd, Switzerland) recorded from 8 mussels: serratus anterior, middle deltoid, pectoralis major, upper trapezius, biceps brachii, triceps brachii, rectus abdominis, latissimus dorsi. The five experimental conditions include manipulating hand medial-lateral location (different medial-lateral distance between hands). The medial-lateral conditions consisted of five hand distance: 0, 20, 40, 60, 80 cm. Raw EMG signals were full-wave rectified and linear enveloped (4 order, low-pass, Butterworth, filter at 4 Hz). Peak EMG was averaged over the 3 repetitions; averaged signals was then described as a percentage of MVIC. The average EMG expressed changes in EMG amplitude over time and was determined by the evaluation of the mean area under the electromyography curve. Statistical analysis was completed using one-way ANOVA for each muscle were used to identify differences in both the mean and peak values. Where post hoc testing using Bonferroni multiple analysis was determined to evaluate the significant differences. Alpha was set to $P \leqslant 0.05$. Also, paired t-tests were determined for each condition and muscle.

Demonstration

Figure 1 presents, peak and mean of 8 muscle activations for five types of hand medial-lateral locations during push-up exercise. Results present differences in the muscle activity amplitude among different hand medial-lateral location. The main effect of medial-lateral hand location influenced all muscles ($p < 0.05$) except upper trapezius and biceps brachii. A medial hand location increased muscle activity (MVC) in four out of eight muscles ($p < 0.05$) during the push-up cycle.

Figure 1 shows, lateral hand location increased muscle activity of the serratus anterior muscle ($p < 0.05$). The major changes happened in the pectoralis major between 25% and 75% MVC.

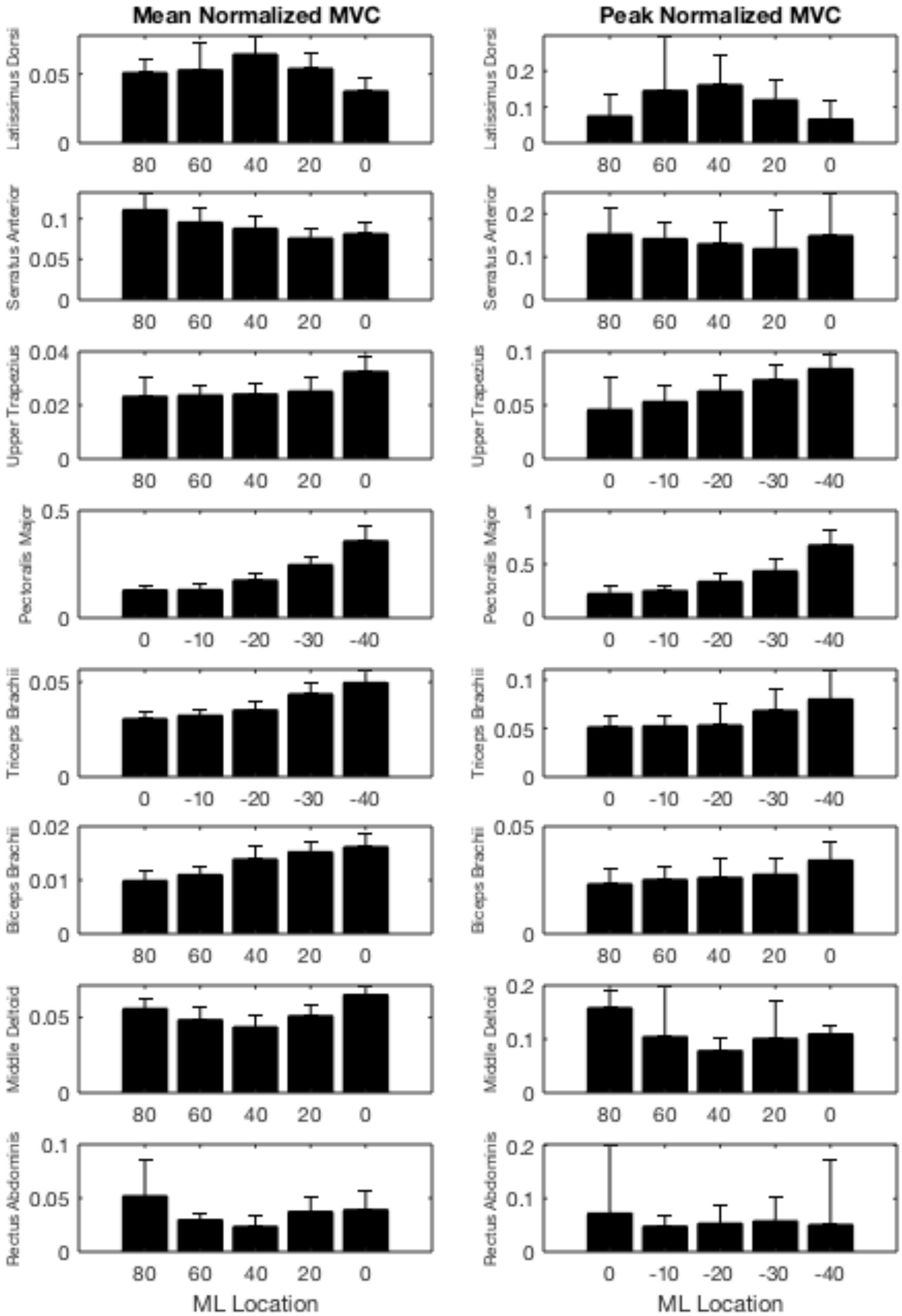

Figure 1: Mean and peak MVC values from five medial-lateral hand locations during push up exercise for eight muscles: serratus anterior, middle deltoid, pectoralis major, upper trapezius, biceps brachii, triceps brachii, rectus abdominis, latissimus dorsi.

Summary

The focus of the current study was to evaluate neuromuscular activation changes when changing hand medial-lateral during the push-up exercise. Medial change of hand location decreased muscle activation in the serratus anterior, while lateral hand location decreased muscle activation pectoralis major, upper trapezius, biceps brachii, and triceps brachii muscles. In addition, medial-lateral hand location in rectus abdominis, latissimus dorsi, and middle deltoid muscles include increasing and decreasing pattern of muscle activations. These changes indicate that medial-lateral hand locations may also promote strength adaptation of the aforementioned muscle. Based on results it appears all five push-ups will evoke equal capability conformity. Neuromuscular activation changes in different medial-lateral hand locations can potentially help athletics, coaches, personal trainer, and clinicians to utilize the modified push up exercises to get new and useful exercise aims that train specific muscles or muscle groups.

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Investigation of loading characteristics in anterior cruciate ligament of target-side knee during professional golf swing

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INTRODUCTION

The combination joint kinematics and kinetics during golf swing suggested that the soft tissue structure of the knee resisting joint compression and internal rotation at low flexion angle about 0-30° may be susceptible for target-side knee injury. Therefore, an anterior cruciate ligament (ACL) rupture is one of the potential injuries that may occur from excessive and repeated stress resulting from the golf swing. In our previous study [1], we estimated the maximum ACL force of the target-side knee during golf swing as 841.8 ± 437.1 N and the result revealed that fatigue life of ACL could be several thousand (minimum of 3000 cycles). However, our study also demonstrated that the inter-subject variation of the maximum ACL load (standard deviation ± 437N) among ten professional golfers was high. This indicates there is a player-specific ACL loading characteristics exist for golf swing motion.

CLINICAL SIGNIFICANCE

The current study is aimed to investigate biomechanical factors leading to high ACL load of the target-side knee during golf swing based on the motion capture data. Our hypothesis is that there are biomechanical factors, which enlarge ACL force during golf swing, and those can be observed from kinematics and kinetic data of golf swing motion.

METHODS

Ten professional golfers were divided into two groups based on the estimated maximum ACL forces (Table 1) in our previous study [1]. Low ACL group includes 6 subjects with ACL force ranging from 270 N to 842.6 N and high ACL group consists of 4 subjects with ACL force ranging from 1069.7N to 1504.1 N (Table 1). The primary interested parameters that compared between high and low ACL groups were lower limb kinematics (hip, knee, ankle joint angles of the target side leg), joint angles of the lower trunk, kinetics of the lead knee (net joint force and moment), ground reaction force (GRF), the external knee moment as well as moment arm (the moment or moment arm generated by GRF relative to lead knee), and trunk-pelvis separation angle or X-factor.

RESULTS

There were notable differences between low and high ACL groups for the magnitude of lower trunk extension (about 6°) and X-factor (38°) at high ACL loading phase. Differences between groups were also observed for the result of GRF moment arm on the frontal plane (about 5 cm) and external knee adduction moment (about 0.3 Nm/BW) of target –side leg.

Table 1: The high and low ACL group classification based on the estimated maximum ACL force in our previous study

	Subject #	ACL force	ACL force / BW	% of swing phase at impact	% of swing phase at max ACL force
low	subject1	663.4	1.07	71	75
high	**subject2**	**1360.9**	**1.84**	**62.0**	**67**
low	subject3	471.0	0.76	58	63
low	subject4	647.3	0.83	69	69
low	subject5	335.7	0.51	61	67
high	**subject6**	**1069.7**	**1.34**	**64.0**	**76**
low	subject7	842.6	0.94	68	73
low	subject8	270.0	0.47	57	57
high	**subject9**	**1504.1**	**1.90**	**62**	**69**
high	**subject10**	**1253.3**	**1.55**	**67**	**67**

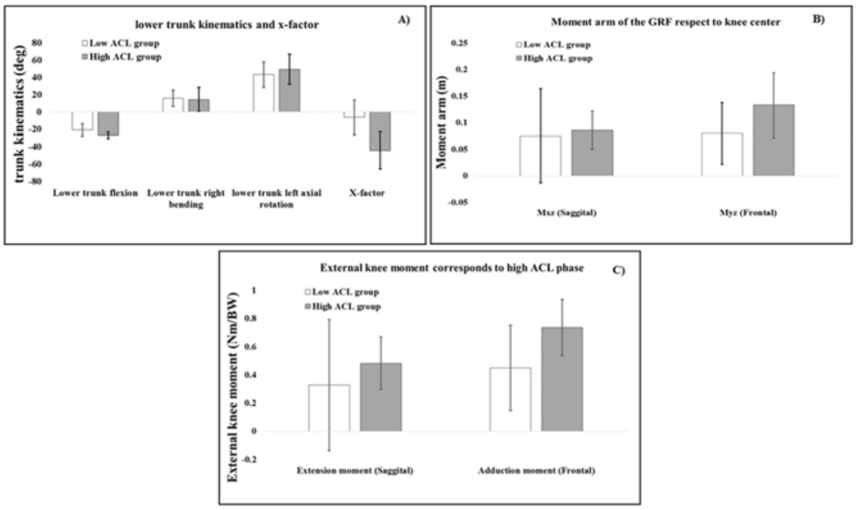

Figure 1: Comparison of the trunk kinematics (A), GRF moment arm respect to target-side knee (B), and external knee moments at high ACL loading phase of golf swing between low and high ACL groups

DISCUSSION

The results at the high ACL loading phase demonstrated an increased amount of frontal plane moment arm and external knee adduction or varus moments just after ball impact, those associated with characteristical difference in the upper body motion or posture, were the main contributors to the elevated ACL force of the target-side knee.

ACKNOWLEDGMENTS

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DISCLOSURE STATEMENT
All authors have nothing to disclose.

Upper extremity prosthetic selection influences loading of transhumeral osseointegrated systems

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***Introduction**:* The abandonment rate of prosthetic devices in transhumeral amputees is as high as 40%, with "too much fuss" being the number one reason for rejection [1]. Recent initiatives to create an upper extremity prosthesis with capabilities similar to a natural limb have enabled amputees to closely achieve pre-amputation function. In parallel, research on percutaneous osseointegrated (OI) endoprostheses for prosthetic attachment has rapidly progressed. By eliminating sockets, OI increases range of motion and humeral rotation that was previously impaired. These advancements allow amputees the possibility of accomplishing more complex motions and improving daily independence. Previous research examined loads on a transhumeral OI system for high demand activities using non-amputee motion to estimate forces and moments on the transhumeral amputation site. Though this gave a better idea of how a high performing individual with transhumeral amputation loads an OI device, it does not look at the impact of a prosthesis on loads during the same activities. In this study, we revisited the data of Drew and Izykowski *et al*. [2] and modified subject models to reflect the mass distribution of an upper extremity prosthesis after OI endoprosthesis implantation. Models reflecting 4 levels of prosthetic limb complexity, from body powered devices to the state-of-the-art advanced prosthetics, were applied to the previous motion capture data and allowed us to quantify the loads experienced at the simulated bone-implant interface in a transhumeral OI system.

***Clinical Significance**:* Gradual loading of OI implants is recommended during early rehabilitation to ensure bone ingrowth [3], but overloading can damage ingrowth as it forms [4-7]. These data provide clinicians developing post-operative rehabilitation protocols an estimate of how the choice of prosthesis changes the load experienced at the transhumeral OI attachment.

***Methods**:* Upper extremity motion capture data from Drew and Izykowski *et al*. [2] was used for the present study. This included 40 non-amputee subjects performing six activities: jumping jack, jug lift, underhand toss, jogging, rapid internal rotation, and briefcase carry. Virtual amputations were created at 25, 50, and 75% humeral length. A single prosthetic arm model was created in Visual3D for each of four categories of limbs based on hand type as a surrogate for overall complexity: body powered, myoelectric hook, myoelectric hand, and advanced prosthetic limb. Elbow width, wrist width, hand length, forearm mass, hand mass, forearm center of mass, and hand center of mass were altered for each prosthetic arm category. A pylon connection was implemented for all models. All prosthetic arm models were applied to each subject's motion data for all three virtual amputation lengths (**Fig 1**).

We examined loading estimations (bending and torsional moments and axial forces) in the same manner as Drew and Izykowski *et al.* [2] for the intact model and each prosthetic model (body powered, myoelectric hook, myoelectric hand and advanced prosthetic) at different amputation levels (25%, 50% and 75%, **Fig 1**). We then employed the multivariate regression approach to simultaneously model force measurements. To assess the differences, we estimated the percentage difference in the force measurements (1) between each prosthetic model to the intact model, (2) between prosthetic models, and (3) between amputation lengths. All statistical comparisons were performed using statistical software R at an *a priori* significance of 0.05.

***Results**:* For all activities that did not have a weight in hand, the body powered prosthesis decreased bending moments 67-87% (range of means for each activity including the 25%, 50%, and 75% lengths), torsional moments 85-88%, and axial pullout forces 60-70% compared to the intact loading values ($p \leq 0.001$). The myoelectric hand model showed the most overall similarity to the intact model with a decrease in bending moment of 1-11%, torsional moment 3-56%, and axial pullout force -1-25% (axial pullout force increased at the 75% amputation level for jogging). The advanced prosthetic increased bending moments 77-101%, torsional moments 64-226%, and axial pullout forces 33-85% ($p \leq 0.001$). Prosthetic models had a smaller impact on change in force from the intact model during activities with a weight in hand.

***Discussion**:* These results reveal a ranked order in loading magnitude according to complexity of the prosthetic device. When comparing the results of this study to those of initial stability failure load measurements on transhumeral OI systems [8], we see axial failure occurs at 4.8x maximum estimated forces, but there is overlap between bending and torsional failure and estimated moments. This overlap highlights the need to further constrain loading and employ protective measures during the early post-operative period when failure occurs at lower loads. Also, it highlights the fact that prosthetics used are an added component to this risk and should be considered when prescribing rehabilitation. OI increases range of motion and advanced prosthetics increase functionality opening the possibility of performing higher demand activities that impart more force on the bone, but this also increases the risk for fracture for individuals with OI attachment systems.

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***Disclosure Statement**:* We have nothing to disclose.

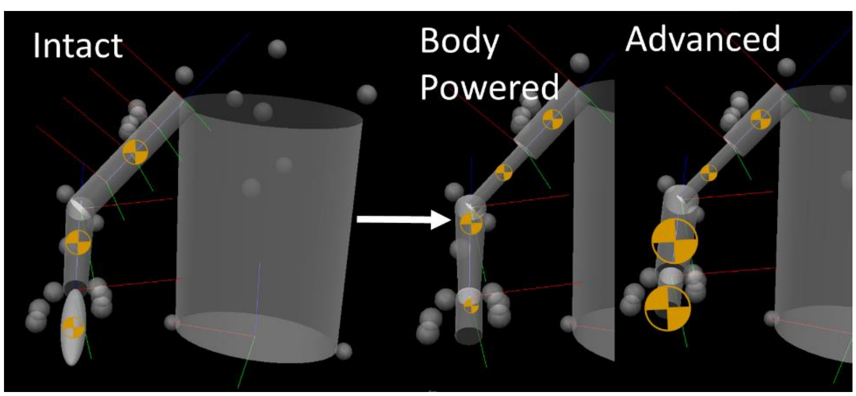

Fig 1. Model modification from intact (left) to body powered and advanced prosthetic model (right). A representative 50% amputation level is shown, but 25%, 50%, and 75% amputation levels were created. ⊕ indicates center of mass and is scaled for segment mass magnitude.

Velocity and position dependency of the stretch reflex in adults with post-stroke spasticity- a neuromechanical approach to assess spasticity

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Introduction

Spasticity has been defined as a motor disorder characterized by a velocity dependent increase in tonic stretch reflexes (muscle tone) with exaggerated tendon jerks resulting from hyperexcitability of the stretch reflex as a component of the upper motor neuron syndrome"[1]. Often seen in children and adults with neurological conditions such as cerebral palsy and stroke, spasticity is caused by an imbalance between excitatory and inhibitory inputs to alpha motor neurons. Spasticity is usually assessed clinically by rating resistance to passive movement (RTPM), by means of the Ashworth and/or Tardieu Scales. However, in addition to stretch reflex activity, RTPM is influenced by non-neural factors, such as biomechanical changes within the muscle [2] due to excess of collagen in the extracellular matrix [3]. Furthermore, due to the anatomy and physiology of the muscle spindle, in addition to the velocity of the stretch, the stretch reflex activity is also influenced by the initial muscle length [4]. This work aims to demonstrate the feasibility of eliciting and characterizing stretch reflex responses at different velocities and at different initial muscle lengths and propose temporal and magnitude parameters as outcomes for the evaluation of the stretch reflex excitability.

Clinical Significance

Existing clinical outcome measures of spasticity are unable to dissociate between altered biomechanical properties of the muscle and the neurological input from the stretch reflex. This can lead to inadequate spasticity treatment if RTPM is due to muscle tissue property changes and not due to increased neural activity. The technique proposed in this work can be used to quantitatively analyze stretch reflex activity and contribute to determine whether increased RTPM is due to increased neural activity, biomechanical changes, or a combination of both.

Methods

A spring driven biomechanical device was designed to stretch the biceps muscle within 100ms at 5 different velocities by applying different torque intensities (2.28 Nm, 3.42 Nm, 4.56Nm, 5.70Nm and 6.84 Nm) at two different initial joint positions (initially flexed, 120 degrees flexion and initially extended, 90 degrees flexion). The electrical activity resulting from the stretch reflex was then recorded using surface electromyography. The stretch reflex signal was then characterized by amplitude, time delay, rise time and duration (figure 1). A general linear model was used to investigate any interaction effects between the initial joint position and the applied angular velocity on the stretch reflex parameters.

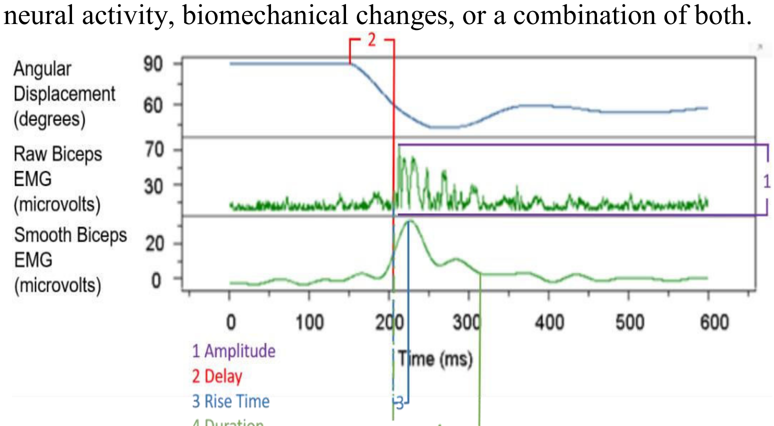

Figure 1

Demonstration
Seven post stroke volunteers who were clinically diagnosed with and treated for upper limb spasticity were included in this feasibility study. Stretch reflexes from the biceps were elicited for all subjects at different stretch velocities and two different initial starting joint angles. Results from the general linear model (figure 2) suggest that the stretch reflex delay and duration were influenced by the initially applied torque and consequently angular velocity, whereas the rise time and duration were dependent on the initial joint position ($p<0.05$). There was no clear effect on the amplitude, though a tendency for it to increase was observed in the initially extended position at higher initially applied torques.

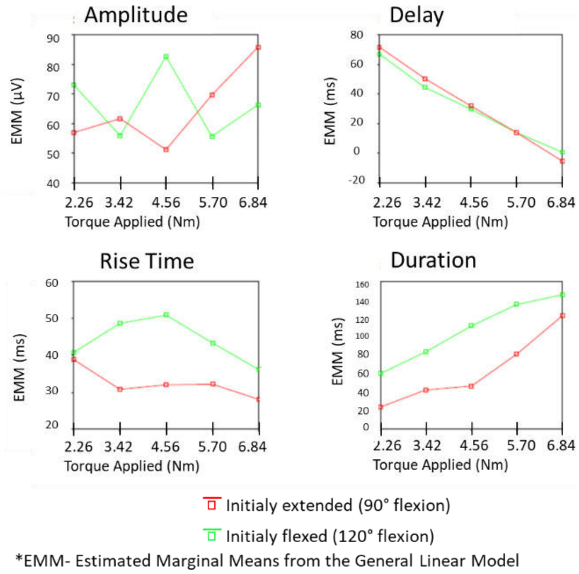

Figure 2

Summary
Spasticity is a complex phenomenon often associated with impaired upper and lower limb motor control. Clinically available techniques, such as the Ashworth Scores and the Tardieu Scale are unable to differentiate between components of joint stiffness. Furthermore, the changes in the stiffness encountered will change depending on the properties of the muscles and initial muscle length of testing. The technique proposed in this work has the potential to characterize and quantify stretch reflex activity in response to standardized and controlled stretch, minimizing the influence of confounding factors. Preliminary data from this work suggests that stretch reflex parameters can be influenced by the velocity of the stretch and the initial joint position, thus they need to be controlled for during assessment.

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Disclosure Statement
No relevant or material financial interests that relate to the research described in this abstract.

INSTRUMENTED WALKER FULLBODY KINETICS IN SUBJECTS WITH CEREBRAL PALSY

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INTRODUCTION

In gait analysis, if a patient requires the use of assistive devices, kinetic calculations usually cannot be made due to the patient putting an unknown amount of load through the devices. This project attempts to solve that problem by instrumenting assistive devices with load cells. Described here is a method of instrumenting a walker with 6-axis load cells, creating a new full body model, and preliminary results comparing lower extremity kinetics in patients with cerebral palsy (CP) who use a walker or walk independently. We hypothesize that patients who use a walker will have decreased lower extremity kinetics compared to patients who walk independently.

CLINICAL SIGNIFICANCE

Using the described hardware and software techniques, kinetic calculations can be made in patients who require assistive devices, generating valuable information not usually available.

METHODS

Two AMTI 6-axis load cells with handles were fixed to a Crocodile walker using custom machined adapters (Figure 1A). The load cells were connected to the data collection computer, via AMTI GEN5 amplifiers, and synchronized with the rest of the data streams in Vicon Nexus.

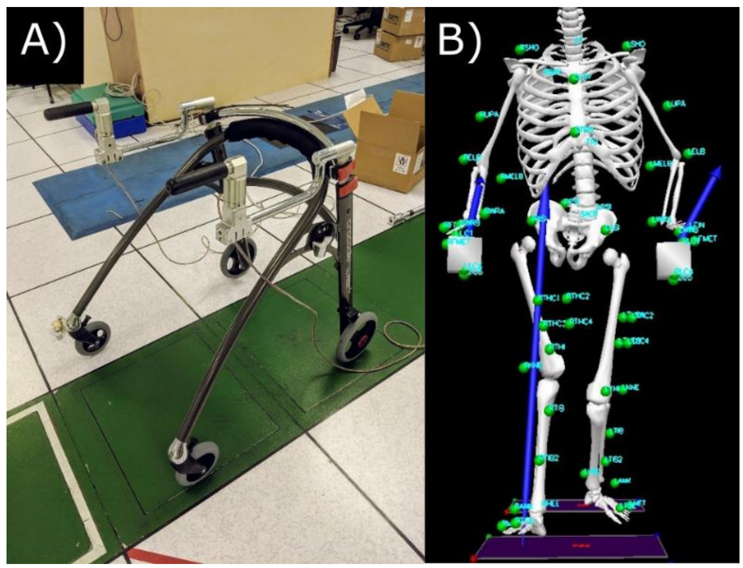

Figure 1. A) Walker customized with load cells B) Full body marker set, hand and ground reaction loads

A custom full body marker set was used, based on Plug-in-Gait (PiG) and 6-degree-of-freedom (6DOF) marker sets. The data was collected in Vicon Nexus and then imported into Visual3D for modelling. In Visual3D a full body 6DOF model was applied; the load cell forces were moved to the load cell origin and applied to the hands (Figure 1B). Moments were normalized to subject body mass.

DEMONSTRATION

The data of three subjects who used a walker (all male, aged 9, 10, and 40yrs) and three subjects who walked independently (all male, aged 9, 9, and 45) were analyzed. The instrumented walker handle height was set to the same height as the subject's walker or to where they felt most comfortable. The full body marker set was placed on the subjects and they walked back and forth across the lab at a self-selected pace while data was collected. No subjects who used the walker made consecutive force plate strikes, therefore only the kinetics of the single-limb-stance phase (SLS) was plotted (Figure 2). Lower extremity moments in subjects who used a walker were not noticeably greater or less than the subjects who walked independently. At the beginning of SLS, knee and ankle sagittal plane moments were greater in independently walking subjects, whereas hip sagittal plane moments were greater in subjects using a walker.

SUMMARY

Using instrumented assistive devices enables collection of full body kinetics in subjects who do not walk independently, although there are many challenges, including handling cables, keeping the walker off the force plates, not providing additional support to the subject, using additional markers, and trying to get the instrumented assistive device to match what the subject is comfortable using. Overall, we did not observe any notable difference in normalized kinetic magnitude between the two groups. However, the largest differences in moments were observed at the beginning of SLS. This may be related to offloading weight onto the walker (reduced ankle and knee moments) and needing to push it forward (increased hip moments) during this phase of gait. Further analysis and an increased sample size are needed to better answer these questions.

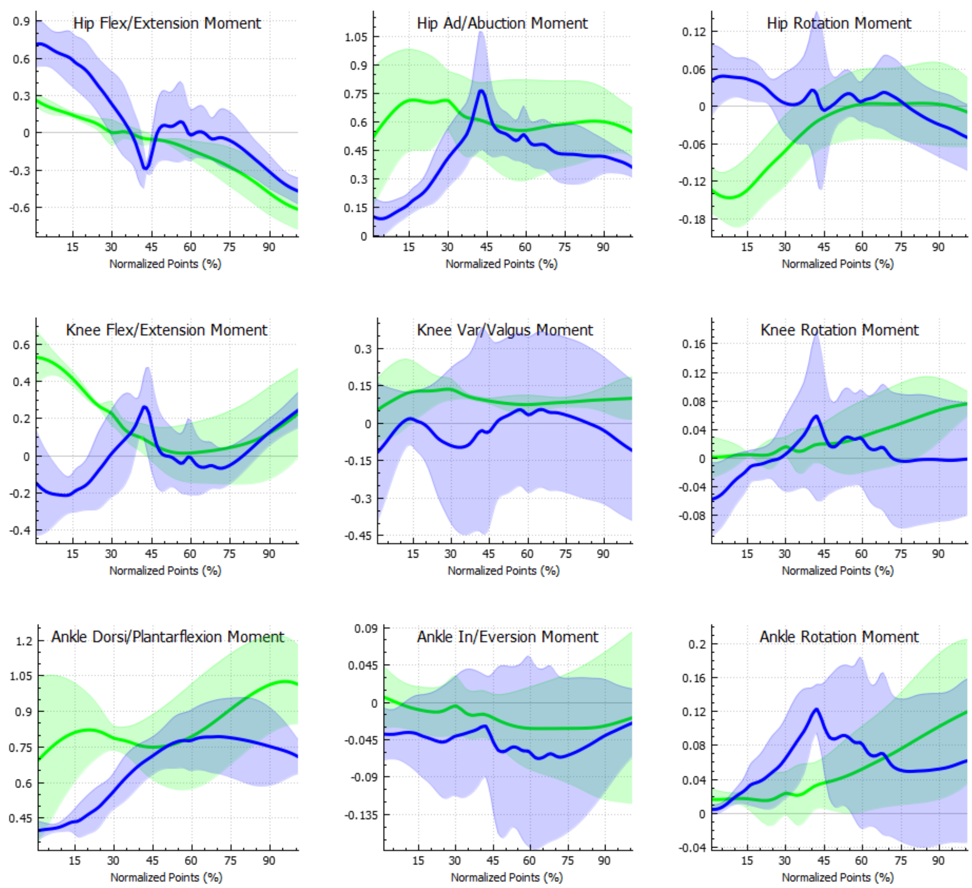

Figure 2. Average (+/- 1 SD) joint moments from SLS for three CP subjects using an instrumented walker (Blue) and three CP subjects walking independently (Green).

DISCLOSURE STATEMENT

The authors have no conflicts of interest to disclose.

Development of an Assistive ExoSkeleton to Improve Walking and Treatment Outcomes in Individuals Recovering from Diabetic Foot Ulcers

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INTRODUCTION

Globally, a lower limb is lost to diabetes approximately every 20 seconds[1]. In the United States, diabetic foot ulcers (DFUs) are the leading cause of lower limb amputations and are now a more expensive burden to the US healthcare system than lung cancer[2]. Despite the potentially severe consequences of having a DFU, DFUs are still common with up to 15% of individuals with diabetes experiencing a DFU during their lifetime[3]. Regardless of severity, a DFU limits mobility and consequently reduces quality of life[4].

Both magnitude and frequency of foot loading influence the prevention and healing of plantar DFUs: international clinical guidelines specify that biomechanical offloading is a critical component of any effective treatment strategy[1]. The transition time period that occurs after wound closure but before total recovery is as critical as the acute phase: individuals return to activity but may have abnormal foot loading patterns or gait deviations (e.g., due to muscle atrophy) that predispose them to re-ulceration.

The aim of this study was to determine how a novel offloading device with a passive exo-tendon aimed at improving clinical and gait outcomes alters basic gait variables (e.g., stride length) and plantar forces during this transition period. This device, called FlyBand®, uses an external spring that can be loaded during stance to provide a supplementary plantar flexion moment during pre-swing (Figure 1A). Through this mechanism, the device aims to offload clinically significant pressure in areas at high risk for re-ulceration (i.e., metatarsal heads) while also assisting to reduce gait deviations due to muscle atrophy.

CLINICAL SIGNIFICANCE

The newly developed FlyBand® has the potential to be a critical component in current programs that support healing and prevention of re-ulceration in individuals that have previously experienced a diabetic foot ulcer (DFU) by improving care during the transition period that occurs immediately after a DFU heals.

METHODS

Fifteen (15) participants (9 male, 6 female) were recruited for this study. Participants had either experienced a diabetes-related ulcer on the plantar surface of the foot in the past or were considered "at-risk" for developing a DFU. To determine the ability of the FlyBand® to improve gait metrics and offload metatarsal plantar force, participants were asked to walk in a straight line (10 m) at a self-selected speed while wearing the FlyBand® without spring tension (i.e., shoes only) or set at one of three tension settings (low, medium, high).

During each walking trial, gait metrics were collected using a LEGSys system consisting of two inertial measurement sensors (one for each shank; BioSensics, LLC, USA). Foot pressure was measured using a sensor placed under the insole of a participant's right foot (Tekscan, Inc., USA). Data were collected at 100 Hz. Stride length, walking speed, and stance phase mean and peak force (calculated from the pressure readings located within the foot's metatarsal area) were calculated and averaged across strides. A repeated measures ANOVA was used to test for significant differences between the different footwear conditions.

Table 1: Measured Stride and Metatarsal Force Variables at Each Walking Condition

Variable (Mean ±1SE)	**FlyBand® No Spring**	**FlyBand® Low Spring**	**FlyBand® Medium Spring**	**FlyBand® High Spring**
Stride Length (m)	1.16 ± 0.05	1.16 ± 0.05	1.20 ± 0.05	1.20 ± 0.05
Walking Speed (m/s)	1.0 ± 0.06	1.02 ± 0.07	1.08 ± 0.06	1.10 ± 0.07
Mean Force (N)	241 ± 22	200 ± 19	198 ± 20	208 ± 23
Peak Force (N)	650 ± 56	570 ± 60	576 ± 64	572 ± 66

RESULTS

Average and peak plantar forces were reduced while using the FlyBand® at all spring tension settings, with the greatest reduction occurring in mean force at the medium spring setting (18%, Figure 1B). Stride velocity and stride length increased in the medium and high spring conditions (Table 1).

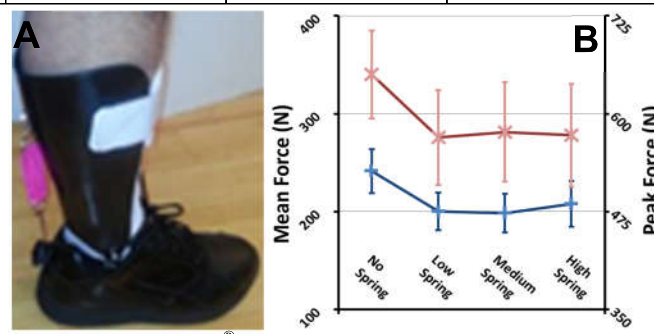

Figure 1: A) FlyBand® device. Springs (pink) provide tension to generate a plantar flexion moment. B) Average (Blue +) and peak (Red *) metatarsal forces versus walking condition (±1SD).

DISCUSSION

The FlyBand® device shows significant promise in improving clinical outcomes for individuals at risk of re-ulceration by providing offloading and improved gait metrics during the critical transition time period that occurs after DFU closure. In individuals with diabetic foot pathologies, the device simultaneously increased walking speed and reduced metatarsal force values, where re-ulceration rates are highest. These benefits were most pronounced in the medium spring condition, suggesting that there exists an optimal level for assisting individuals. Future work will determine how these gait and mechanical changes directly relate to clinical outcomes such as re-ulceration rate.

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DISCLOSURE STATEMENT

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A New Hip Joint Center Predictive Method for Children with Developmental Dysplasia

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INTRODUCTION

During gait analysis it is important to accurately determine the joint centers of the lower extremities as they are utilized in the computation of the lower body kinematics. The calculation of the thigh segmental reference frame is based on the location of the hip joint center (HJC) therefore the kinematics of the hip (thigh relative to pelvis) and knee (shank relative to thigh) joints are directly dependant on the location of the HJC.

HJC offsets as little as 6 mm have been shown to cause a significant change in mean hip and knee flexion and extension, as well as, mean hip and knee abduction and adduction angles[1]. Therefore, the accurate estimation of the HJC location is vital in assuring correct kinematics during gait analysis. To the authors' knowledge, there has not been a predictive method for estimating the HJC created for patients with developmental dysplasia of the hip (DDH). Therefore, the purpose of this study was to create a new method to estimate the HJC that can be used for patients with hip dysplasia.

CLINICAL SIGNIFICANCE

This new method will more accurately compute hip and knee kinematics and kinetics in people with hip dysplasia.

METHODS

Twenty-eight patients (24 female, aged 18.7 ± 4.1 yrs) diagnosed with unilateral or bilateral DDH were enrolled into an institutional review board approved study were identified for this study. Patients were excluded if they had secondary diagnoses such as cerebral palsy, Charcot-Marie-Tooth or Down syndrome. Weightbearing bilateral anteriorposterior (AP) pelvic radiographs were measured for each patient to determine the medial-lateral and superior-inferior location of the HJC relative to the anterior superior iliac spine (ASIS), as well as the lateral center edge angle (LCEA)[2], acetabular inclination (AI) and the extrusion index (EI)[3]. Since an AP film only shows the frontal plane of the pelvis, the HJC position in the sagittal and transverse plane were calculated using the Davis method.

A new regression equation using predictive variables such as LCEA, ER, AI, pelvic depth (PD), pelvic width (PW), and leg length (LL) was created for the DDH cohort. Stepwise regression analysis was completed to determine which of the predictive variables has the strongest impact on the HJC location in the medial-lateral direction (HJCx). A custom MATLAB (MathWorks, Natick, MA, USA) code was written to compute the lower body kinematics, and was modified to include 5 HJC calculation methods: 1) Davis[4], 2) Harrington[5], 3) new Radiographic based method, as described above, 4) Functional method[6], 5) True position derived from the ASIS to HJC (mediolateral) coordinates measured on xrays. The orientation angles of the hip and knee joints were computed during gait and static trials. A comparison of the effect of the deviation in predictive methods for the HJC was assessed using paired t-tests.

RESULTS

The results of the stepwise regression analysis showed that PW, LCEA, and EI were significant predictors of HJCx while PD and LL were poor predictors of HJCx. Using the three strongest predictor variables, a new HJC predictive method regression equation was created:

$HJCx = 0.351(LCEA) - 0.181(EI) + 0.121(PW) - 2.861$ where LCEA is the lateral center-edge angle in degrees, EI is the extrusion index in percent, and PW is pelvic width in millimeters.

Static Comparison: The medial-lateral HJC locations of the four alternative methods were compared to the true position of the HJCx location based on the radiographic measurements (Table 1). The average position of the Radiographic based method was closest to the True position (-6.41mm) compared to Davis (-16.29mm), Harrington (-9.85mm), and Functional (15.19mm). Overall, the Davis method (medially) and the Functional method (laterally) were the poorest predictors of the HJC in the mediolateral position.

Gait Comparison: No significant differences were seen in the sagittal and transverse planes. In the coronal plane (Table 1), the new Radiographic method performed the best with average hip abduction/adduction and average knee varus/valgus through the gait cycle all within 1° of the kinematics computed using the true HJCx position.

Table 1. Differerences between the four methods used and the true position measured from AP xrays of the mediolateral hip joint center position from the four methods used and the true position measured from AP radiographic images during static and gait trials.

	Static Comparison			**Gait Comparison**					
	HJCx Position			**Hip Abd/Add Avg**			**Knee Var/Val Avg**		
Method	***Mean (mm)***	***SD (mm)***	***p-Value***	***Mean (°)***	***SD (°)***	***p-Value***	***Mean (°)***	***SD (°)***	***p-value***
Radiographic	-6.4	9.6	**<0.001**	-0.9	4.1	0.148	0.8	4.2	0.213
Davis	-16.9	11.2	**<0.001**	-2.4	4.2	**<0.001**	2.2	4.4	**0.002**
Harrington	-9.9	10.6	**<0.001**	-1.4	4.0	**0.003**	1.3	4.3	0.055
Functional	15.1	13.6	**<0.001**	2.2	4.4	**0.021**	-1.8	4.7	**0.018**

DISCUSSION

The purpose of this study was to develop a new regression-based prediction method for the medial-lateral position of the HJC which utilized radiographic measures in dysplastic hips. Applying this method to dysplastic hips, the HJC was lateralized compared to the previously published Davis and Harrington methods. Although the radiographic-based method computed a HJC location that was the closest to the true HJCx position in normal hips, the Harrington method was shown to be more accurate than the Davis or SCORE method, which agrees with previous research[5]. This study assumed that the center of the femoral head was the center of rotation of the hip . It is unclear if this is the case in DDH patients which do not have typical morphology of the pelvic and hip. One limitation to this study is that only the ML direction was assessed. Future work should assess the tri-planar position of the HJC. In the future, advanced imaging (e.g., MRI, CT, ultrasound) could be used to determine the true HJC location in three-dimensions. The regression equations derived from these measurements would then be able to describe the location of the HJC relative to the ASIS in all three planes, providing a more accurate representation of the HJC position.

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DISCLOSURE STATEMENT: All authors have no conflicts of interests to disclose.

A COMPARISON OF METHODS TO ESTIMATE SCAPULAR KINEMATICS FOR BRACHIAL PLEXUS BIRTH INJURIES

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INTRODUCTION

Glenohumeral (GH) joint deformity, dysplasia and range of motion (ROM) deficits as well as scapulothoracic (ST) compensations are commonly observed in patients with brachial plexus birth injury (BPBI) [1,2]. Various interventions are employed to improve GH function, however, assessment of GH function in dynamic conditions is lacking due to challenges in measuring dynamic scapular orientation and the inability to distinguish between GH and ST contributions to shoulder motion. The acromion marker cluster with single calibration (S-AMC) has been commonly used to estimate dynamic ST orientation in healthy populations [3] but was deemed unsuitable for BPBI [4]. An updated AMC method using double calibration (D-AMC) to enable interpolation of estimated ST orientations based on humerothoracic (HT) elevation [5] is currently recommended [6] due to its substantially improved accuracy over the S-AMC, but has yet to be assessed in children with BPBI. Additionally, a linear model (LM) approach utilizing measurable HT orientation and acromion process position to estimate ST orientation has been validated in healthy adults [7], but only preliminarily assessed for BPBI over a limited shoulder ROM [8]. This study evaluates the abilities of the S-AMC, D-AMC, and LM to estimate ST orientation in static positions across a full shoulder ROM by comparing against palpation. We hypothesized that ST errors between the LM and palpation would be less than those of D-AMC and much less than those of the S-AMC.

CLINICAL SIGNIFICANCE

Accurate measurement of ST motion is needed for objective evaluation of dynamic ST and GH joint function, which will facilitate surgical planning and outcomes assessment in BPBI.

METHODS

Nineteen children with BPBI participated. Trunk and arm orientations were measured using ten markers, an AMC, and a 12-camera motion capture system (VICON, Centennial, CO). Subjects completed 14 static positions encompassing the full spectrum of shoulder ROM. In each position, two additional scapular markers were placed to determine ST orientation. HT angles were calculated with a helical approach and an Euler YXY sequence. An Euler YXZ sequence was used to calculate ST angles. All 14 positions served as test positions. For each position a "leave one out" approach was used to develop a LM using multiple linear regression on data from the 13 other positions to generate equations that estimated ST angles based on HT orientation (helical XYZ and Euler Y") and acromion process position. ST angles estimated by S-AMC, D-AMC, and LM were compared to palpation and average root mean square errors (RMSE) for each approach were calculated across all participants on each ST axis of motion for each test position.

RESULTS

LM and D-AMC produced comparable RMSE values that were generally within established acceptable limits (<10°) [3] for most positions/axes; S-AMC values were at times much larger (**Fig. 1**). The percentage of errors >10° for all subjects and positions in this study varied across the three ST axes: LM (9.9% - 16.0%), D-AMC (1.8% - 7.3%), S-AMC (2.9% - 9.2%).

DISCUSSION

Estimating ST orientation in BPBI is more challenging than in a healthy population. RMSE values for all methods were close to the upper range or beyond those of previous healthy analyses [3]. LM had exceptionally large errors for certain positions indicating that they involve HT or ST angles beyond those of the other 13 positions used to develop the LM. This highlights a LM limitation that should be considered when using this approach. The LM's greater likelihood of producing >10° errors for an individual subject showed that this method is less reliable than D-AMC for assessing BPBI. Similar to past findings [5], D-AMC was superior to S-AMC especially on internal/external rotation and anterior/posterior tilt. Given the generally acceptable errors and ease of use, the D-AMC may be suitable for BPBI in certain scenarios.

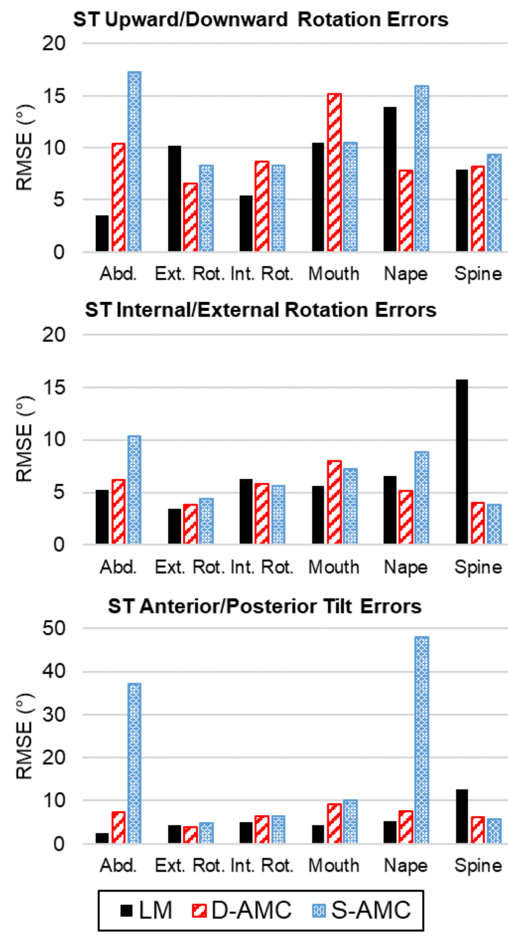

Figure 1: RMSE values for ST orientation at selected test positions.

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DISCLOSURE STATEMENT

The authors have no conflicts of interest to disclose.

PRELIMINARY ASSESSMENT OF A TOOL TO MEASURE UPPER EXTREMITY WORKSPACE USING REAL-TIME FEEDBACK WITH MOTION CAPTURE

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INTRODUCTION

Clinical upper extremity (UE) functional assessments have been criticized for being insensitive to certain meaningful differences in UE function [1]. More precise motion capture measurements are typically constrained to a set of static postures and/or limited dynamic motions relevant to the patient population [2] and may provide an incomplete assessment of a patient's available UE functionality. Similar to clinical assessments, motion capture analyses are also reliant on patient compliance with traditional task-based data collection protocols both of which can be challenging to collect on younger patients.

Assessing UE functionality by utilizing a less prescriptive and more game-like environment may provide the opportunity to obtain precise measures of UE function on young children [3]. The measurement of reachable workspace can be obtained in such an environment. Reachable workspace provides a precise numerical and visual assessment of global UE function by quantifying the surface area or volume that the patient can reach with his or her hand [4] and represents an emerging tool among UE surgeons [5]. Incorporating real-time motion capture visual feedback into current UE workspace measures can provide an innovative way to engage patients [3] while also ensuring acquisition of the patient's entire available workspace.

This study provides a preliminary evaluation of a new clinical tool that incorporates real-time visual feedback with motion capture to quantify UE reachable workspace.

CLINICAL SIGNIFICANCE

A reachable workspace tool with real-time feedback may provide a more complete measure of global UE function, facilitate testing in younger patients, and provide a highly visual and intuitive depiction of patient function for clinicians, patients, and caretakers.

METHODS

One healthy adult participated. Trunk and UE segment orientations were measured using 13 markers and a 10-camera motion capture system (Qualisys, Göteborg, Sweden). An array of virtual targets was created using spherical coordinates where r = distance from origin (sternum), θ = angle of elevation, and φ = plane of elevation. The target radii were scaled to arm length. Custom LabVIEW software (National Instruments, Austin, TX) displayed targets with real-time feedback from motion capture (**Fig. 1**). Movements of a red sphere were controlled in real-time according to the position of the participant's hand marker relative to the trunk coordinate system (**Fig. 1**). Once the red sphere moved within a threshold of a target, that target would

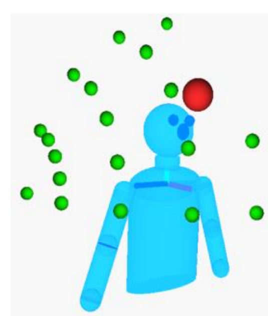

Figure 1: Real-time feedback display. Green targets and red real-time sphere.

disappear. Targets throughout the spherical space were displayed sequentially by octant (e.g. upper hemisphere, right side, anterior). Targets in each octant were displayed for 30 seconds or until all targets were completed leading to a maximum trial time of four minutes. For this preliminary proof-of-concept the participant completed two trials with his right arm: 1) normal healthy UE function, 2) simulated impaired UE function.

All hand coordinates that were >70% of maximum radius reached were analyzed. Reachable workspace space was calculated as the points reached during the trial expressed as a percentage of the potential reachable points along the outer surface area of each octant.

DEMONSTRATION

Healthy UE workspace was greater than simulated impaired workspace (**Table 1**). Visualizations of the healthy and impaired workspaces showed these differences and facilitated easy identification of regions with the greatest impairments (**Fig. 2**).

Table 1: Percentage of each octant reached.

Upper Hemisphere		
Octant	Healthy	Impaired
R. Anterior	88.9%	44.4%
L. Anterior	66.7%	22.2%
R. Posterior	43.2%	8.6%
L. Posterior	8.6%	0.0%
Lower Hemisphere		
Octant	Healthy	Impaired
R. Anterior	86.4%	95.1%
L. Anterior	56.8%	27.2%
R. Posterior	46.9%	38.3%
L. Posterior	6.2%	0.0%

SUMMARY

This preliminary proof-of-concept demonstrated the viability of a reachable workspace tool with real-time feedback to provide an alternative measure of global UE function that is capable of clearly illustrating areas of impairment. Further exploration and refinement of this tool in healthy and clinical populations is warranted.

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ACKNOWLEDGMENTS

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DISCLOSURE STATEMENT

The authors have no conflicts of interest to disclose.

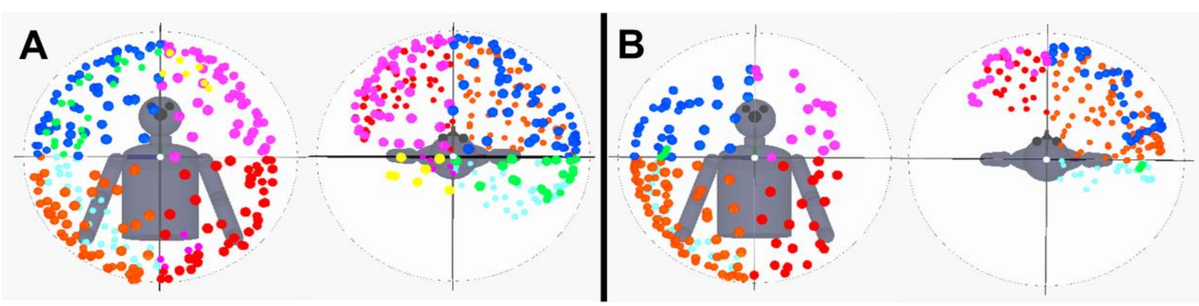

Figure 2: Frontal and superior views of UE workspace of healthy (A) and simulated impaired (B).

Proof of Concept: Step and Stride Time Calculation using an Inexpensive Inertial Measurement Unit for Future Assessment of Gait Symmetry in Hemophilia

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Introduction

Laboratory-based motion analysis is the gold standard for quantifying gait performance and identifying gait pathology. However, typical instrumented gait analysis (IGA) in a laboratory setting is expensive, not widely available in clinics or in the field, and overkill for simple screening of temporal-spatial asymmetry, especially for patients with non-neuromuscular diagnoses. We have developed a technique to calculate step and stride time using two miniature $90 inertial measurement units (IMUs) attached to both feet, that may offer an alternative for simple screening of gait pathology. This project presents results from a preliminary validation study where sensor-based data are compared to step and stride times obtained from a conventional clinical motion analysis facility.

Clinical Significance

Hemophilia is a bleeding disorder in which clotting factor deficiencies lead to joint and muscle bleeding. Persons with hemophilia (PwH) can have alterations in gait related to pain, swelling, or arthritis from joint bleeding.[1,2] Having a simple, inexpensive tool to measure timing of gait events bilaterally could enhance the care of these individuals and provide a simple quantitative measure of disease progression.

Methods

After providing informed consent, full 3D kinematic and kinetic data were collected from 10 healthy adult volunteers at the UCHealth Foot and Ankle Center Motion Analysis laboratory along with simultaneously collected IMU data from both feet. Each participant walked at a comfortable speed for several passes with IGA data recorded using a Vicon® motion capture system employing 8 Vantage cameras (120 Hz) and 6 Bertec force platforms (2160 Hz). MbientLab® MetaMotionR IMUs were attached to each participant's shoelaces on both feet using clips. Measured triaxial accelerations were stored to the device and collected over Bluetooth using MbientLab software at 100 Hz. Foot/floor timing events were identified using Vicon Nexus 2.0, and custom Matlab software extracted and synchronized the IMU data for further analysis. Stride times were calculated comparing time of initial foot strike for subsequent impacts with the same foot, using both Vicon and IMU data. Left step time was defined as time from left weight acceptance to end of left single limb support using data from both IMUs, and right step time was defined using the same events but referenced to the right side. For step time IMU calculations, IMUs were synchronized using a prominent signal from a drop jump. Stride times from both systems were calculated from 3-5 representative cycles referenced from both right and left limbs for 10 participants, and from all valid steps from one trial (approximately 30 steps) from one participant. Step times were also compared for all valid steps in that participant.

IMU- and Vicon-calculated stride and step times were analyzed using intraclass correlation (ICC) two-way mixed effect models for consistency. To evaluate between subject consistency of Vicon and IMU samples, average stride times measured using 3-5 steps from 10 participants were compared using intraclass correlation (ICC) two-way mixed effects model of absolute agreement.

Results/Demonstration

Figure 1 shows characteristic acceleration signals from shoe IMUs during walking. <u>Single participant evaluation:</u> Stride times measured using Vicon vs IMU showed an ICC of 0.80, calculated either from left foot impacts or right foot impacts. Comparison of left step times measured using Vicon vs IMU yielded an ICC of 0.89, and comparison of right step times yielded an ICC of 0.86. <u>Group evaluation:</u> The average stride times calculated using Vicon vs IMU for 3-5 strides in 10 participants (Figure 2) yielded an ICC of 0.94 for steps calculated from left foot impacts and 0.93 from steps calculated from right foot impacts.

Figure 1: Characteristic acceleration signal from IMUs clipped to shoelaces. Triangles represent timepoints used for initial impact.

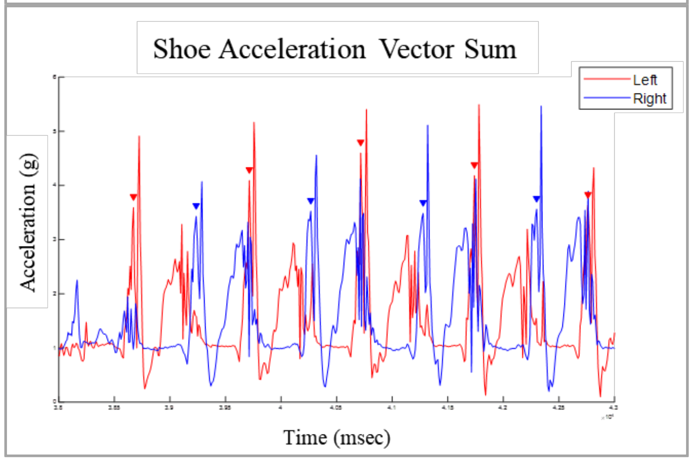
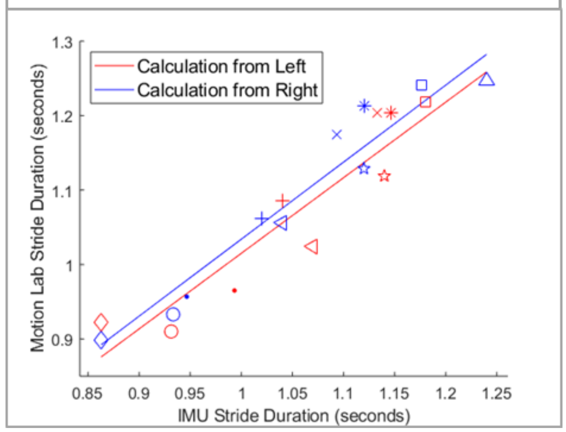


Figure 2: Relationship between stride times calculated from Vicon and from IMUs. Each symbol represents a different participant.



Summary

ICCs between Vicon- and IMU-calculated data indicated that IMUs were able to measure step and stride times comparable to motion capture, suggesting that IMUs could be a good surrogate for temporal-spatial event detection and symmetry calculation. When evaluating between-subject measurement consistency, ICCs between Vicon- and IMU-calculated data showed excellent agreement, suggesting that both methods are consistent across individuals with different walking speeds.

Analysis of additional participants is needed to demonstrate wider technique applicability, including the inter-individual consistency of step time. Future analysis will also include root mean square error, although improved techniques, likely involving continuous video, will be needed to link specific steps measured with Vicon to IMU signals. These data serve as a proof of concept for collecting gait event data from extremely low cost IMUs.

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Acknowledgements

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Disclosures

The authors have no relevant conflicts of interest to declare.

JOINT CONTACT FORCE MODEL FOR PATIENTS WITH KNEE HEIGHT ASYMMETRY

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INTRODUCTION

Limb length discrepancy (LLD) is a common condition, affecting 40-70% of the population [1], [2]. LLD can be corrected through shortening, lengthening, and/or growth arrest, depending on the extent of the discrepancies and whether or not angular deformities are present [3], [4]. During these procedures, however, symmetry between corresponding long bones is usually not prioritized, leading to knee height asymmetry [5]. Cadaveric studies have shown that knee height asymmetry is associated with early onset osteoarthritic changes [6], which could be a result of atypical loading during repetitive activities such as walking. To date, there is minimal information on what effect knee height asymmetry has on gait biomechanics and joint loading following limb lengthening surgery. Computational modelling can be used to estimate joint loading during everyday activities. Thus, the purpose of this study is to use a computational model of the lower limbs to determine whether knee height asymmetry causes atypical loading in the knee joints during gait.

CLINICAL SIGNIFICANCE

The ability to calculate joint contact forces in patients with knee height asymmetry will help determine whether knee height asymmetry results in atypical joint loads. With this knowledge, surgeons will then be better equipped to establish standards for the surgical treatment of LLD.

METHODS

Mass and height for a lower limb OpenSim model [7], [8] with 23 degrees of freedom and 92 musculotendon actuators were scaled in OpenSim using marker position data collected during a static calibration trial. Subject kinematics were calculated using marker-based inverse kinematics calculations. Ground reaction forces from four force plates and the kinematics were input into a residual reduction algorithm to reduce force and moment residuals by altering patient pelvic mass and center of mass location so that the model kinematics match the kinetics more closely until the residuals were near the software recommended ranges. The static optimization algorithm was used to solve for individual muscle forces at each time point by constraining muscle force and length properties and minimizing the muscle activations. Knee joint reaction forces were calculated using OpenSim's JointReaction analysis, a method detailed by Steele et al. [9].

DEMONSTRATION

Table 1: Subject Demographics

	KHA subject	Control 1	Control 2
Sex	Male	Male	Female
Age	17 years	16 years	12 years
Height	164.2 cm	171.6 cm	159.4 cm
Weight	105.2 lbs	134.5 lbs	126.3 lbs

Data was collected using the Vicon Plug-in gait lower body model (Vicon Nexus, Vicon Motion Systems Ltd., United Kingdom) for one subject with knee height asymmetry and two healthy controls, both with legs of equal length and symmetrical knee height (Table 1). The subject with knee height asymmetry is post-operative for LLD correction, initially with an external fixator (tibial) from 2015-2016, and then using the PRECICE nail (NuVasive®, Inc., San Diego, CA) in the right femur from 2017-2019. He achieved full limb lengthening, with bilateral lower extremities within equal range of each other, but his right limb is now 3 cm longer than the left, resulting in knee height asymmetry (femur L/R 48/50 cm; tibia L/R 36/37 cm). Knee joint contact forces were calculated for up to 3 trials for both the left and right limbs; mean contact force curves were compared between limbs and between subjects (Figure 1).

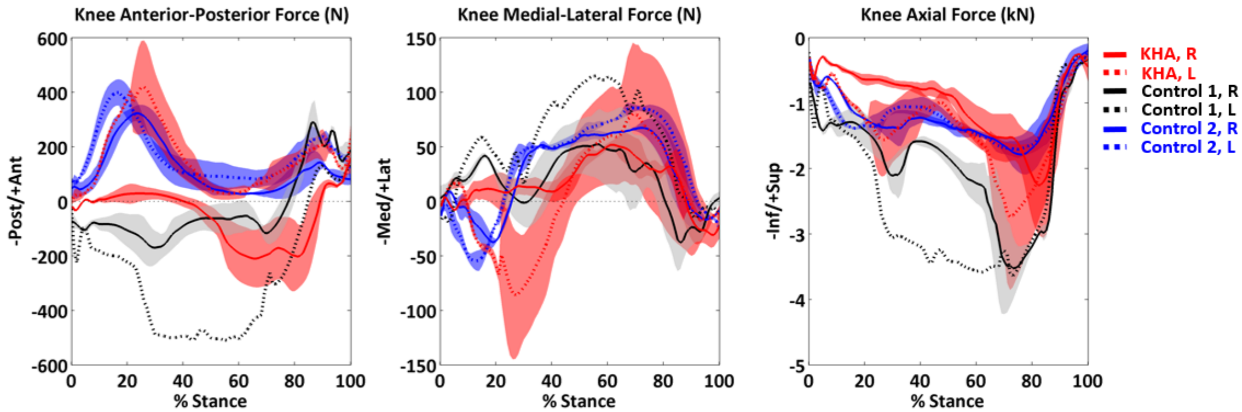

Figure 1: Right and left knee three-dimensional joint contact forces compared between a subject with knee height asymmetry (KHA) and two control subjects. The curves show the mean force ± 1 standard deviation of all trials from each limb. Solid lines indicate right side trials; dashed lines indicate left side.

SUMMARY

This abstract demonstrates the feasibility of using musculoskeletal modelling to analyze knee joint contact forces in patients with knee height asymmetry. Results are within a feasible range, as they are similar in magnitude to knee joint contact forces measured by telemetric total knee replacements [10]. They show different patterns of anteroposterior force in the knee joint between the right and left sides compared to the more symmetrical patterns shown by the second control subject. This research demonstrates a technique which can be used to analyze joint contact forces in patients with knee height asymmetry for a variety of activities of daily living. Future studies will look at joint contact forces for patients with knee height asymmetry to determine any atypical loading which could indicate a need to adjust the surgical management of children with LLD.

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DISCLOSURE STATEMENT - We have no conflicts of interest to disclose.

Gait and Functional Outcomes of Adults with Cerebral Palsy

M. Wade Shrader, M.D., Chris Church, M.P.T., Nancy Lennon, M.S., P.T., Faith Kalisperis, M.P.T., John Henley, Ph.D. , and Freeman Miller, M.D.

<u>**Purpose**</u>: Advances in pediatric orthopedic care over the past three decades have improved mobility, function, and quality of life for children and youth with cerebral palsy (CP). The long-term effectiveness of this care into adulthood has not been widely reported. Some studies demonstrate gait outcomes in young adults with cerebral palsy (CP) that show walking mechanics are largely unchanged from adolescence, but the relationship of gait outcomes to participation and pain in adulthood is unknown. Walking with persistent gait deviations increases energy cost, produces fatigue, and may lead to joint pain and degenerative joint disease in the long-term.

This purpose of this course is to review the relevant literature of gait and functional outcomes of adults with cerebral palsy, to discuss the difficulties and challenges of long-term follow-up studies in this population, and to describe in detail a recent investigation that measures long-term outcomes of outcomes of physical function, mental well-being, participation and pain in a cohort of adults with CP who received specialized pediatric orthopedic care as children and adolescents.

<u>**Course Format:**</u>
Introduction, Overview of Course Objectives:
M. Wade Shrader, M.D. (5 minutes)

Review of Previous Literature of Gait and Functional Outcomes of Adults with CP:
Faith Kalisperis, M.P.T. (15 minutes)

- Long-term gait follow-up studies
- Longitudinal assessment of function in adults with CP
- Identification of current gaps of knowledge

Design of a Call-Back Study: Who do our pediatric patients grow up to be?:
John Henley, Ph.D. (10 min)

- Purpose and need for long-term follow up study
- Study design
- Logistics of adult patient follow-up, contacting, and recruitment

Gait and Functional Outcomes of 130 adults with CP followed for 13 years
Chris Church, M.P.T. (15 min)

- Discussion of Methods and Subjects
- Adult kinematics and kinetics compared to adolescence
- Can we predict in adolescence who will decline in function as adults?

Pain in Adults with Cerebral Palsy: Review of the Literature and Results of our Call-Back Study
M. Wade Shrader, M.D. (10 min)

- Review of current literature of adults with CP
- Does high quality pediatric orthopedic care impact the prevalence of pain in adults with CP?

Functional Outcomes: Patient Reported Quality of Life and Activity
Nancy Lennon, M.S., P.T. (15 min)

- PROMIS patient reported outcomes of pain, participation, and life-satisfaction
- Community activity monitoring
- Implications for further study

Case Examples of Expected Adult Outcomes
Freeman Miller, M.D. (20 min)

- Interactive case review

Question & Answer and Discussion Panel: All speakers (30 minutes)

Target Audience: Physicians, Occupational and Physical Therapists, Engineers

Prerequisite Knowledge: Basic knowledge of quantitative gait analysis and pathophysiology of cerebral palsy

Learning Objective: After completion of this course, participants will be able to explain the current level of evidence of long-term outcomes of gait and physical function and their correlations with patient reported outcomes of mental well-being, participation, and pain in adults in cerebral palsy.

Simple Uninstrumented Tools (Video Based 2D Motion Analysis and Forceline Visualization) to Optimize and Quantify Prosthetic Gait

Cara Negri, BSME, MS, CP, FAAOP
Mukul Talaty, PhD

Purpose

While Instrumented Gait Analysis (IGA) is the accepted gold standard, each session can take up to six hours for assessment and interpretation (Narayanan, 2007). IGA is not a practical outcome measure for many clinicians. Clinicians require simple and cost-effective outcome measures to analyze the kinematic and kinetic parameters of gait in their day-to-day practice (Rathinam 2014). VGA in conjunction with other carefully selected outcome measures – like EVGS and POGS – can provide a comprehensive gait assessment in situations where when IGA is not available or not indicatedpractical (Harvey & Gorter, 2011).“Good quality video as part of a 2D motion analysis can result in data that are comparable to that of instrumented motion analysis systems when used for 2D analysis, enhancing clinician’s evidence- based practice.” (Fatone & Stone 2015). There is an increasingly important role for Visual Gait Analysis (VGA). Structured gait scoring techniques, like Edinburgh Visual Gait Score (EVGS) and Prosthetic Observational Gait Score (POGS), can lead to more objective and reliable methods than what is common practice today in Orthotics and Prosthetics. The EVGS and POGS are (two of) the most reliable observational gait assessment tools available to clinicians. Prosthetic alignment can affect safety, functional ability, energy cost, comfort and aesthetics of gait. Though there is no universal definition for an optimal alignment, there is a group of characteristics that is loosely associated with appropriate or acceptable alignment – especially for unilateral transtibial amputees. Dynamic forceline visualization (DFV) is a way to visualize without having to guess the location of the line of force relative to lower extremity joints. It is inexpensive, does not encumber the patient, and provides immediate semi-quantitative estimates of gait kinetics. When combined with suitable guidance, DFV can be used to get immediate feedback on the consequences of modifications to prosthetic or orthotic alignment and thus can be used to make improvements to alignment to facilitate better walking outcomes. We will share some empirically derived alignment targets – a loose set of guidelines based on normal gait biomechanics. These, along with the biomechanical relationships of how alignment changes in specific componentry can influence gait kinetics will then be applied to guide changes to prosthetic alignment.

Intended Audience

Members of a healthcare or research team who are interested in applying objective Gait Analysis methods – especially those of gait involving prosthetics and orthotics – but who may not have state of the art gait lab resources available.

Prerequisite Knowledge
None. Participants will benefit more if they have familiarity with the basic gait cycle, biomechanics of normal gait and have some experience with observational gait analysis.

Outline:

1. Introduction
    1. The role, limitations, and advantages of video and observational gait analysis.
    2. The realistic clinical setting
2. 2D Video and the Prosthetic Observational Gait Score
    1. Precautions
    2. Best Practices- Review of the research (Fatone and Stine, 2015)
    3. Uses- Case studies
3. Demonstration- Using an amputee patient model, perform a 2D Video assessment and complete the POGS assessment.
4. [10 mins] Introduce the concept of forceline – background, examples of different types and available resources, provides an estimate of joint kinetics
5. [10 mins] Discuss how can use to evaluate frontal & sagittal planes at various times in gait cycle – using "normal" targets [Talaty M, Esquenazi A. Determination of dynamic prosthetic alignment using forceline visualization. Journal of Prosthetics and Orthotics. 2013;25:15–21. doi: 10.1097/JPO.0b013e31827afc29]
6. [20 mins] Demonstration of Application of "Alignment Targets" – per above - using Amputee model. Can we improve this alignment?
7. [15] mins Post-Alignment POGS reassessment. What did we change?
8. [20 mins] Questions and Discussion
9. Using the Score to Plan for Care

Learning Objectives
At the completion of this presentation, attendees will better understand how simple tools (video and forceline) can be use to objectively measure and modify prosthetic gait.

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DISCLOSURE STATEMENT
Cara Negri is the owner of PnO Data Solutions which distributes The Tarn Group Products for using video analysis. However, this presentation will be unbranded to show the use of the score and video and not to promote the system.

Establishing Early Mobilization Rehabilitation Guidelines Following Single Event Multi-level Surgery with the Assistance of Motion Analysis Technology

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Introduction

Early mobilization is a key component of our rehabilitation program at Shriners Hospitals for Children in Houston for our ambulatory children with Cerebral Palsy who have undergone a Single Event Multi-level Surgery (SEMLS). A recent survey of the literature and various SEMLS programs revealed that institutions vary widely in protocols for initiation of weight bearing, with many delaying weight bearing for 6 to 8 weeks. The Shriners Houston team approach for the past decade has been to initiate weight bearing within the first 48 hours post-op. This more aggressive philosophy allows the patient to return to school and ambulation in all settings within 4-6 weeks post-op. Prior to this change in treatment philosophy, these patients at our facility would have typically remained non-weight bearing for 6-8 weeks. With the assistance of motion laboratory technology and analysis, we have been able to provide evidence to support the benefits of early mobilization in this patient population. This is an example of how motion analysis technology can be used to guide clinical practice and provide objective data for evidence based medicine and improved patient outcomes.

This course will cover a wide range of SEMLS topics from assessment to rehabilitation to outcomes analysis. We will highlight how the outcomes in our Motion Analysis Center have shaped and influenced our post-operative rehabilitation guidelines and patient selection. A discussion of guidelines for intensity and length of stay for rehabilitation will be included.

Clinical Significance

The aim of this course is to foster discussion and share knowledge concerning early mobilization in this patient population.

Methods

Instructional strategies will include lecture and gait videos with PowerPoint presentation. Knowledge translation will occur by incorporating didactic information into case studies throughout the tutorial. There will be time allotted for open dialogue with guided questions as needed.

Discussion

Historically post-SEMLS rehabilitation required a lengthy immobilization period post-operatively along with delayed weight bearing, leading to increased caregiver burden [1,3,4]. Upon hospital discharge some patients required a hospital bed, wheelchair, and homebound school services [1,4,6,7]. Decreasing post-op immobilization is crucial, as research has found that children with spastic CP are weaker than typically developing children [3]. In many institutions, epidurals are used for post-operative pain control, which delays weight bearing and ambulation. Shriner's intensive early mobilization and ambulation program combined with multi-modal pain management and rigid internal fixation of osteotomies allows restoration of participation in daily activities, ambulation and school within the first month post-op. Using data from our motion analysis laboratory has helped us fine tune our rehabilitation guidelines and improve outcomes.

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Disclosures: There are no conflicts of interest to disclose.

Title: Consistent Interpretation of Gait Analysis Data: A Case-Based Quality Assurance Approach

Instructors: Tom Novacheck, MD, Andrew G. Georgiadis, MD, Jean Stout, PT, MS

Purpose: The purpose of this course is to demonstrate a quality improvement process used to assess consistency gait interpretation. Participants will have opportunity for "hands on" experience interpretation consistency process for problem identification and treatment recommendations with other participants. Gait analysis case studies in conjunction with an audience response system/survey monkey system will be employed.

Intended Audience: Pediatric orthopaedists, physical therapists, physical medicine and rehabilitation physicians, biomechanics, kinesiologists, engineers, and other professionals who participate in the interpretation of gait analysis data.

Prerequisite Knowledge: Intermediate. Gait analysis interpretation skills are used.

Course Summary: This course will summarize our experience with the design and success of an interpreter consistency quality assurance program and certification process. Use of an audience response system/mobile technology will allow participants to be interactive in assessing their own consistency among other participants and responses from Gillette Children's interpretation staff. A series of case studies will be used to illustrate problem identification and treatment recommendations. The controversies surrounding gait interpretation consistency within and across centers and its impact on the utility of gait analysis will be discussed.

Course Format:

Introductions and Cases	5 minutes	Stout
Controversies Surrounding Interpretation Consistency of Gait Analysis and Recent Evidence	10 minutes	Georgiadis
Design of an Interpreter Consistency and Certification QI program	15 minutes	Novacheck
Case 1: Audience Participation of Problem Identification	20 minutes	All
Questions & Answers	10 minutes	All
Break	5 minutes	
Review Study Data: Problem Identification MD & PTs	10 minutes	Stout
Case 2: Audience Participation: Problem Identification & Treatment Recommendations	25 minutes	All
Review Study Data: Treatment Recommendations	10 minutes	Georgiadis
Wrap Up, Questions & Answers	10 minutes	All

Title:

Lessons Learned: How to Design a Multi-Center Motion Analysis Study

Instructor(s):

Kirsten Tulchin-Francis, PhD[1], Sherry Backus, PT, DPT, MA[2], Danilo Catelli, PhD[3], Marcie Harris-Hayes. PT, PhD[4]. Mario Lamontagne PhD[3], Cara Lewis PT, PhD[5], David Podeszwa, MD[1]

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Part of the Motion Analysis in FAI Study Group, supported by the Pediatric Orthopedic Society of North America

Purpose:

To provide an overview of the planning involved in the development of a multi-center, multi-task motion analysis protocol.

Intended Audience:

Any individual involved in motion analysis (orthopedists, sports medicine specialists, biomechanists, physical therapists, athletic trainers, kinesiologists and engineers) who may consider participating in the functional evaluation of participants in a multi-center study group or project.

Prerequisite Knowledge:

No particular prerequisite knowledge is required.

Abstract:

Supported by a Clinical Planning Grant from the Pediatric Society of North America, our study group aimed to develop a multi-task, multi-center motion analysis protocol for the assessment of patients with femoroacetabular impingement (FAI) syndrome. This grant was not to conduct the study, but rather, to fund the development of such a multi-center trial. Grant activities included multiple conference calls, in-person planning meetings, the development of a comprehensive standard operating procedure, and pilot testing with 5 healthy control adults without FAI morphology.

The aim of this tutorial is to share our experience and "lessons learned" from this process, and to provide basic guidelines and best practices that can be followed to develop comprehensive standard operating procedures for multi-center research in motion analysis.

Learning objectives:

At the completion of this course, attendees will:

- Understand the importance of standard operating procedures for multi-center motion analysis projects

- Discuss the role that pre-project planning meetings, demonstrations, group discussion and other documentation have in the development of multi-center motion analysis project standard operating procedures
- Understand the clinical and technical content that should be included in multi-center motion analysis project standard operating procedures
- Discuss the role of pilot testing for assessing reliability across centers
- Discuss the variables of interest and selected methods to assess data reliability

Outline of course content:

I. Introduction to course (10 mins)

This will include the course outline and learning objectives as well as background on why multi-center motion analysis studies are needed.

II. You have a project idea and interested/committed collaborators, how do you start to develop a new, standardized, multi-center protocol? (15 mins)

This section will review the information that was gathered on the front side, before discussions on the protocol even began. Once gathered, how can these data be discussed and aggregated in a productive manner to move the project forward?

III. Data collection and processing standardization (60 mins)

Understanding the importance of using strict guidelines for each and every part of the protocol is vital. Here we will discuss what steps we took to develop standardization for clinical measures, marker placement and task execution, using a multi-step process.

IV. What is a multi-center motion analysis project Standard Operating Procedure (SOP)? (20min)

The protocol developed will only be as good as its SOP. What information goes into a SOP, and how much detail do you really need to include?

V. Question/Answer (15min)

*Handout will include a sample SOP based on that which was developed for our study group.

Targeted Task Proprioception Protocol Validation Study

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Introduction: Proprioception is an awareness of the position and movement of the body from sensory neurons in the skin, joint tissues, and muscles [1]. Proprioceptive information assists the body with movement through space and aids with the completion of daily activities, such as turning on a light without looking at the switch [2]. In addition to the internal sensory feedback, proprioception is augmented by visual input for the completion of targeting tasks. Prior work has measured proprioceptive accuracy based on remembered visual cues, but these studies constrain motion to isolate individual joints. This joint isolation may not be representative of completing activities of daily living using the upper extremity (UE) [3]. Many techniques for measuring proprioception also use specialized equipment, such as robots to track and drive motion or fixtures to isolate joint use. These methods work well to restrict proprioceptive testing to a single joint; however, most daily tasks require the coordination of proprioception from multiple joints [4]. Therefore, the objective of this study was to validate a tool for assessing proprioception by measuring an individual's performance while completing an UE targeting task. To validate this tool, we measured test-retest reliability via intra-subject variability across trial sessions for each subject.

Clinical Significance: Validation of this tool will provide researchers and clinicians with an assessment that can objectively measure a targeted path of the UE. Our goal was to enable its use as an assessment of functional changes resulting from new prostheses or percutaneous osseointegrated implants in transhumeral amputees. The developed tool is a simple and objective assessment of proprioception that can be replicated objectively with a given participant across sessions during rehabilitation. This tool does not require the use of special equipment, making it accessible for rapid and widespread clinical integration.

Methods: All procedures were approved by the local IRB (University of Utah #00127288), and all participants provided written informed consent. Exclusion criteria include UE limb loss or decreased mobility due to an UE injury. We have completed data collection on 5 participants with anticipated enrollment of 24 subjects, based on our power analysis. Each participant completed two experimental sessions, one week apart to diminish carryover effects. For each session, participants sat with their arms in front of them with elbows shoulder-width apart, at the edge of the table and holding an uncapped marker in each hand using a self selected grip that was consistent for both sessions. A sheet of paper, with an X in the middle of the page, was placed in front of the participant so that the near edge of the paper was aligned with the participant's hands. Each participant was given 5 seconds to look at the X before being instructed to use one fluid motion to make a dot on the page as close to the X as possible. A planar horizontal motion was selected to minimize external feedback and enable the use of multiple joints during the motion [4]. Hand usage and eyes open/closed were randomized and participants completed 40 trials (ten trials in each combination of hand use and eyes open/closed) during each session. For trials with the eyes closed, the marked paper was removed before the participant was instructed to open their eyes to minimize the feedback on their performance. Each trial sheet was scanned and measured using a custom MATLAB script to determine the distance and angle of the drawn dot from the target X.

Demonstration: Demogrpahics for the 5 participants were: median age = 25 (23-29), 5M/0F. All participants were right hand dominant. To assess test-retest repeatability, we conducted a visual analysis of the overlap between the first and second session results (Fig. 1). We observed a substantial overlap in the scatter between sessions; however, due to our limited dataset, we are unable to make statements about the statistical significance of the overlap. For all sessions, the average distance of the dot from the target X was 3.4 ± 2.0 mm and 24.2 ± 11.9 mm with eyes-open and -closed, respectively. There was a difference in accuracy when targeting with the eyes-open versus -closed ($p<0.001$). There was also a distinct grouping of the dots concerning directional bias as represented by the standard deviation of the direction of error angle, where 0° is defined along the positive y-axis. With eyes-open the subjects used the visual feedback to correct their directional error on subsequent trials, resulting in a greater standard deviation in the direction of error angle (Left = 175.9 ± 91.0°, Right = 178.7 ± 79.9°). During the eyes-closed trials, the subjects consistenly undershot the target toward the utilized hand, resulting in a lower standard deviation in the direction of error angle (Left = 78.0 ± 41.6°, Right = 241.9 ± 41.8°).

Summary: Based on the limited sample size, the results demonstrate test-retest repeatability and reveal normative values from a healthy subject population. These trends indicate that outcomes do not change as a result of repetition and support utilizing this study design to create the definitive normal data set. We have developed this tool to represent completion of UE activities of daily living, such as reaching for a light switch or picking up an item from a table, without direct visual cues. Validation will enable future use in limb-loss patient populations to measure changes in proprioceptive performance due to rehabilitation and changes in equipment. In addition to patients with limb loss, this tool could aid in tracking recovery from stroke or serve as a metric in degenerative pathologies such as multiple sclerosis. The establishment of normative values, including accuracy and directional bias relative to the target, strengthens the clinical use of this tool by providing comparisons of proprioceptive performance during rehabilitation, such as left/right symmetry. Future work will focus on assessing the correlation between this tool and daily uses of proprioception to ensure significance for measuring clinical outcomes.

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Disclosure Statement: We have nothing to disclose.

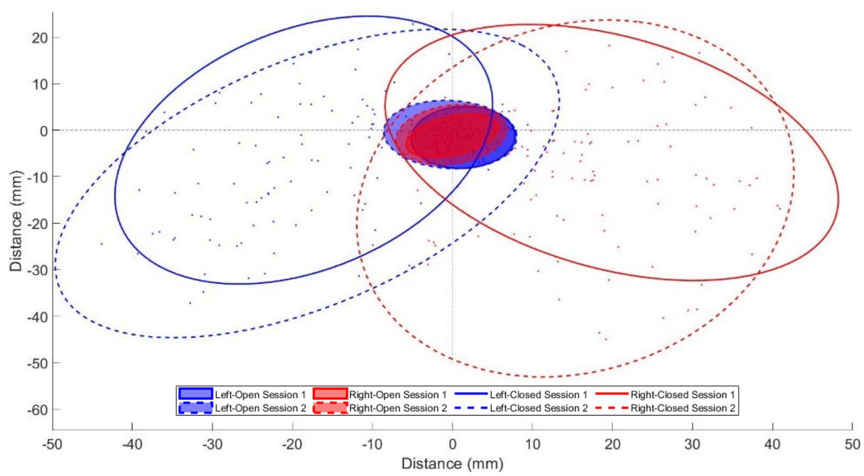

Figure 1 95% confidence ellipses comparing first and second session results for test-retest repeatability through overlap of the confidence ellipses. The major axis of each ellipse is oriented along the direction of greatest variance, and the magnitude of the major and minor axes is proportional to the variance in each direction to determine the ellipse size.

CHANGE OF GAIT AFTER PERIPHERAL DIZZINESS: A PROSPECTIVE 3D MOTION ANALYSIS STUDY

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INTRODUCTION

Dizziness is a frequent symptom from various diseases. Abnormality in vestibular system is one of the most frequent pathologies for dizziness. Peripheral vestibular dysfunction evokes vestibulo-ocular reflex (VOR) deterioration and balance and gait problems. However, VOR adaptation did not go along with recovery gait problems with various severities. Therefore, assessment of gait function after dizziness is essential for relevant treatment. Previous studies reported temporo-spatial parameters which did not have direct relation with dynamic balance during gait. Instantaneous COM-COP inclination angle (IA) provides information about the ability to control COM position in relation to the corresponding COP. The ML COM-COP inclination angle is suggested to be a sensitive parameter of gait stability in older people. We observed IA changes after dizziness onset for 4 weeks and 8 weeks, prospectively.

CLINICAL SIGNIFICANCE

Variability of IA in frontal plane improved after 2 months of dizziness onset. These findings suggest that dynamic stability returned significantly after 2 months of dizziness onset. We believe that clinicians should make more than two months of clinical efforts to prevent falls and restore walking stability.

METHODS

This is a prospective study. This study included participants who visited ENT department for dizziness and were judged peripheral dizziness by ENT specialist. Participants had Vestibular neuritis, Meniere disease, recurrent vestibulopathy or BPPV (benign paroxysmal positional vertigo). Exclusion criteria were 1) comorbidity in CNS such as cerebral infarction, 2) medical history of musculoskeletal problems that could disturb normal walking, such as joint contracture, apparent limb length discrepancy, severe peripheral neuropathy. After diagnosis, standard therapy including medication and education was provided for symptom remission. 3D motion analysis was conducted for level walking after initial diagnosis within 2 weeks. Tests were repeated after 1st study 4 weeks and 8 weeks. During 3D motion analysis, participants walked with self-selected speed. Explicit target was set parallel to the laboratory axis during walking trials to inform the target direction for participants. Fifty-six reflective markers were attached on head, trunk, pelvis, upper & lower arm, thigh, leg and foot segments recommended by Visual3D (C-motion Inc., Rockville, Maryland, USA) Visual3D software was used to calculate gait parameters. Center of pressure (CoP) was calculated from force plate data. Center of mass was calculated from kinematic and anthropometric data. Dynamic stability during walking was assessed by and IA in frontal plane. Variability of IA was calculated with RMSE (root mean square error) in frontal plane. Repeatative ANOVA was used for statistical test with significance <.05.

RESULTS

Of the 122 participants referred for 3D gait analysis study, 37 participants were included in the analysis. Results are reported in Table 1. Walking speed, stride length and both step lengths increased with statistical significance ($p < 0.05$) but not step width. Walking speed improved at 2nd test through increasing stride length. Minimum IA in frontal plane did not show significant changes. Variability of IA in frontal plane reduced 3rd test with statistical significance ($p < 0.05$). Variability of IA showed improvements at 3rd test during gait cycle, single stance phase and double stance phase (Table 1)(Figure 1).

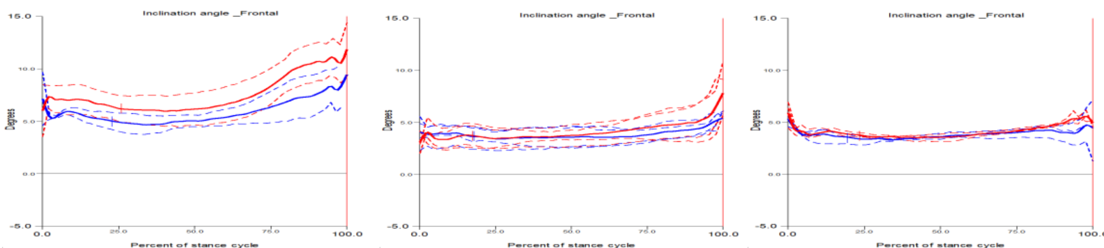

Figure 1. Example case with acute vestibular neuritis, left = 1st, center = 2nd, right = 3rd test.

Table 1: Tables and graphs may span both columns if necessary (mean± SD).

	1st (mean/sd)	2nd (mean/sd)	3rd (mean/sd)	*p*
Speed	1.1293 / 0.1286	1.1602 / 0.1112	1.1641 / 0.0978	**.022***
Stride length	1.2074 / 0.1065	1.2391 / 0.0915	1.2338 / 0.0924	**.022***
Cadence	112.1499 / 7.7403	112.3873 / 7.3915	113.7496 / 8.8508	.165
Step width	0.1089 / 0.0299	0.1060 / 0.0264	0.1096 / 0.0254	.440
Step length right	0.5997 / 0.0561	0.6191 / 0.0476	0.6180 / 0.0475	**.006***
Step length left	0.6074 / 0.0533	0.6198 / 0.0469	0.6213 / 0.0431	**.019***
IAF_min_sum	5.134 / 1.7731	4.7647 / 1.3083	5.1046 / 1.3948	.103
IAF_max_sum	9.9878 / 2.7984	9.8364 / 2.3374	10.1703 / 2.3910	.569
IAF_RMSE	0.5155 / 0.1784	0.5027 / 0.1698	0.3829 / 0.1458	**< .001***
IAF_RMSE_SS	0.4496 / 0.1690	0.4387 / 0.1609	0.3243 / 0.1270	**< .001***
IAF_RMSE_DS	0.6671 / 0.2158	0.6524 / 0.2071	0.5293 / 0.2032	**.003***

IAP; inclination angle in frontal plane. RMSE; root mean square. SS; single support phase. DS; double support phase. * denotes p <.05.

DISCUSSION

Movement variability usually increases after neuromuscular injury and decreases during recovery or adaptation. Observation of medio-lateral variability of CoM-CoP relationship is more relevant than walking speed or step width for gait function recovery in peripheral dizziness.

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DISCLOSURE STATEMENT

Authors have no conflicts of interest to disclose.

GAIT ALTERATIONS ON IRREGULAR SURFACE IN PEOPLE WITH PARKINSON'S DISEASE

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INTRODUCTION

Gait impairments and balance limitations are common among people with Parkinson’s disease (PD). Most PD gait analyses are conducted on smooth, solid surfaces [1]. Few studies have investigated PD gait on irregular terrain due to greater instability and fall risk, yet many individuals with PD commonly experience irregular terrain in daily life. The aim of this study was to characterize gait parameters on irregular surfaces for persons with PD.

CLINICAL SIGNIFICANCE

This information improves the understanding of parkinsonian gait adaptations on irregular surfaces and may guide gait training and rehabilitation interventions for this high fall-risk population.

METHODS

Nine persons with PD (age 67.7±7.1 years, height 1.66±0.16 m, weight 81.0±20.6 kg, UPDRS 36.1±11.8, H&Y 2.39±0.31) and nine healthy age-matched (HA) controls (age 67.7±8.0 years, height 1.69±0.05 m, weight 81.0±20.6 kg) were recruited. Walking trials were collected at self-selected speeds for both PD and HA using a 24-camera motion capture system at 100 Hz. The irregular terrain was simulated using faux rock panels. Persons with PD were in an ‘on’ medication state during the activities. In order to minimize the potential injury, a fall protection system was used.

The spatiotemporal variables (speed, cadence, step length, step width, and single limb support), lower limb kinematic variables (range of motion (RoM) for the hip, knee and ankle joints) and stability variables (trunk RoM and center of mass (CoM) acceleration root mean square (RMS)) were determined [2]. Paired t-test was used to compare between regular and irregular surfaces for both PD and HA groups. The results were considered statistically significant when $P < 0.05$.

RESULTS

The participants with PD showed a decreased speed, cadence, step length and single support, but an increased double support time compared with healthy controls. Both PD and HA groups showed a smaller speed and step length on the irregular surface than on the regular surface. The PD group manifested decreased knee sagittal plane RoM compared to the controls. The effect of group significantly affects the knee sagittal plane RoM (Table 1). A significant surface effect was found for ankle RoM in the transverse plane, which was larger on irregular surface for both PD and HA groups.

The PD group demonstrated smaller trunk RoM in the frontal and transverse planes, but larger mediolateral RMS of trunk acceleration compared with healthy controls (Table 2).

Surface had a significant effect for RMS of trunk acceleration in all three axial directions. A larger RMS of trunk acceleration on the irregular surface compared to the regular surface in all three axial directions was found only for HA group, but not PD group.

Table 1 Kinematic variables expressed as mean (SD) for PD and HA groups

RoM (degree)	Regular Surface		Irregular Surface	
	PD	HA	PD	HA
Hip (sagittal)	39.43 (6.73)	44.08 (4.90)	40.66 (7.24)	45.21 (3.30)
Hip (frontal)	9.32 (2.29)	11.62 (2.34)	9.34 (2.77)	10.66 (1.76)
Hip (transverse)	11.61 (2.27)	11.90 (3.67)	12.10 (3.46)	12.38 (3.03)
Knee (sagittal) †	59.43 (4.67)	67.67 (3.67)	57.35 (7.66)	69.87 (4.15)
Knee (frontal)	10.41 (3.69)	7.96 (3.60)	11.08 (4.59)	8.12 (2.36)
Knee (transverse)	14.09 (2.15)	13.53 (2.88)	13.81 (3.82)	14.68 (3.56)
Ankle (sagittal)	27.42 (4.96)	28.74 (3.61)	26.76 (5.77)	26.79 (2.12)
Ankle (frontal)	12.77 (3.15)	14.09 (3.09)	12.48 (2.68)	13.83 (2.99)
Ankle (transverse) *	11.21 (2.55) #	12.45 (3.56) #	13.59 (3.05)	16.13 (2.13)

* Significant surface effect; † Significant group effect; # Significant difference between two surfaces in PD or HA groups.

Table 2 Stability variables expressed as mean (SD) for PD and HA groups

Parameter	Regular Surface		Irregular Surface	
	PD	HA	PD	HA
Trunk RoM (degree)				
Sagittal plane	4.22 (1.49)	5.05 (1.10)	6.06 (3.85)	7.19 (3.56)
Frontal plane †	7.01 (3.31)	11.10 (2.85)	7.63 (2.86)	11.75 (3.88)
Transverse plane †	12.65 (6.14)	18.47 (6.16)	11.40 (5.16)	19.10 (5.45)
RMS of trunk acceleration				
Anteroposterior *	1.27 (0.42)	1.11 (0.19) #	1.54 (0.40)	1.31 (0.11)
Mediolateral * †	0.88 (0.24) #	0.63 (0.10) #	0.98 (0.24)	0.78 (0.13)
Vertical *	1.76 (0.79)	1.69 (0.40) #	1.85 (0.74)	1.79 (0.49)

* Significant surface effect; † Significant group effect; # Significant difference between two surfaces in PD or HA groups.

DISCUSSION

The irregular surface posed a greater challenge to maintain balance and stability for individuals with Parkinson's disease. A relatively small knee range of motion in the sagittal plane and large root mean square of trunk acceleration increased the potential fall-risk for individuals with Parkinson's disease [3]. Future research may apply these findings to gait training and rehabilitation interventions for persons with PD in a virtual reality environment.

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DISCLOSURE STATEMENT

There are no conflicts of interest to disclose.

ASSESSING FALL RISK THROUGH GAIT ANALYSIS BASED ON INERTIAL SENSORS

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INTRODUCTION: Falls are a major problem for the health of the elderly. Older adults who have experienced a fall are likely to injure their mental as well as physical being, resulting in poor quality of life. Therefore, research to predict the risk of falling is important. Recently, as a fall analysis research, studies have been actively using inertial measurement units (IMU) rather than the motion-capture systems or force plates analyses, because they are not time- or space-consuming and are inexpensive[1]. This study aims to analyze IMU-based gait parameters by analyzing sarcopenia and the history of falls among the relative fall risk factors presented by the American Geriatrics Society (AGS) [2].

CLINICAL SIGNIFICANCE: Gait analysis based on IMU sensors can be used as an evaluation tool to predict fall risk at a low price without consuming time and space in daily life, and without being constrained to the laboratory.

METHODS: The study included data from 60 subjects (20 normal, 20 with history of falls, 20 with sarcopenia) with a mean age of 66 years (standard deviation = 11.28). Data were collected four times in a straight line along the 27-meter corridor. The gait data were acquired from the right and left insole, and the sensor settings had an acceleration sensitivity of 8G, a gyro sensitivity of 1000 ° / sec, and a sampling frequency of 100 Hz. The device is 17x25x3 mm and the processor used is Nordic nRF52840 (ARM® Cortex®-M4 32-bit processor with FPU, 64 MHz), Flash is 512Mbits Flash, Memory is 1 MB flash and 256 kB RAM. The inertial sensor is an Invensense MPU-9250 with 16-bit ADCs, acceleration change sensitivity of ± 2g, ± 4g, ± 8g ± 16g and gyro change sensitivity of ± 250, ± 500, ± 1000, ± 2000 ° / sec. Additionally, it supports Bluetooth® low energy (BLE) mode, external interface for battery charging, real-time data transfer, and data storage. The gait parameters (mean and standard deviation (STD) of cadence, stance time, swing time, single and double support time, and RL_balance (stance right/left)) were assessed to evaluate the risk of falling from the measured data[3]. Three groups (normal, history of falls, sarcopenia) were compared through the detected parameters and a t-test was performed as shown in Table 1. We used SARC-F questionnaire and Morse Fall Scale for group classification.

RESULTS: Subjects (normal, history of falls, sarcopenia) were compared with gait parameters measured by IMU sensors. As a result, sarcopenia patients showed slower walking speed (stance time increases, single support decreases, double support increases, step length decreases, and cadence decreases), history of falls, and decreased RL_balance (the right foot's stance time divided by the left foot's stance time). Additionally, those with a history of falls and sarcopenia showed lowered STD compared to normal participants.

DISCUSSION: As a result of comparing gait parameters using the gait data obtained from the IMU to predict the fall risk, some of the parameters showed significant differences from the major factors (history of falls, sarcopenia) of falling risk compared to the normal group. The study confirmed that by detecting additional gait parameters and applying machine learning techniques, IMU-based gait analysis has the potential to predict fall risk.

Table 1: Analysis of gait parameters between inertial sensor-based fall risk groups

Subject	**Normal, Sarcopenia**		**Normal, History of falls**		**History of falls, Sarcopenia**	
Parameter	**t-score**	**p-value**	**t-score**	**p-value**	**t-score**	**p-value**
Stance time right Mean	-3.23.E+00	1.53E-03	-1.19.E+00	2.35E-01	-1.81.E+00	7.18E-02
Stance time right STD	-3.30.E+00	1.24E-03	-3.84.E+00	1.78E-04	2.89.E-01	7.73E-01
Stance time left Mean	-1.83.E+00	7.09E-02	1.13.E+00	2.62E-01	-2.69.E+00	8.00E-03
Stance time left STD	-3.47.E+00	7.31E-04	-4.93.E+00	2.65E-06	1.45.E+00	1.48E-01
Swing time right Mean	3.23.E+00	1.53E-03	1.19.E+00	2.35E-01	1.81.E+00	7.18E-02
Swing time right STD	-3.30.E+00	1.24E-03	-3.84.E+00	1.78E-04	2.89.E-01	7.73E-01
Swing time left Mean	1.83.E+00	7.09E-02	-1.13.E+00	2.62E-01	2.69.E+00	8.00E-03
Swing time left STD	-3.47.E+00	7.31E-04	-4.93.E+00	2.65E-06	1.45.E+00	1.48E-01
Single support right Mean	1.83.E+00	7.16E-02	-1.24.E+00	2.19E-01	2.82.E+00	5.54E-03
Single support right STD	-3.90.E+00	1.90E-04	-3.77.E+00	2.57E-04	-1.11.E+00	2.71E-01
Single support left Mean	3.16.E+00	1.91E-03	1.02.E+00	3.09E-01	1.86.E+00	6.51E-02
Single support left STD	-3.15.E+00	3.01E-03	-2.26.E+00	2.49E-02	-1.44.E+00	1.53E-01
Double support Mean	-2.71.E+00	7.52E-03	2.64.E-01	7.92E-01	-2.67.E+00	8.37E-03
Double support STD	-1.49.E+00	1.37E-01	-1.68.E+00	9.86E-02	4.70.E-01	6.39E-01
Stride length Mean	4.91.E+00	2.27E-06	5.06.E-01	6.14E-01	4.09.E+00	6.90E-05
Stride length STD	-1.01.E-01	9.19E-01	-1.42.E+00	1.61E-01	1.20.E+00	2.33E-01
Step length right Mean	5.06.E+00	1.16E-06	6.70.E-01	5.04E-01	4.11.E+00	6.59E-05
Step length right STD	4.17.E-01	6.76E-01	-9.95.E-01	3.25E-01	1.29.E+00	1.99E-01
Step length left Mean	4.66.E+00	6.88E-06	3.36.E-01	7.37E-01	4.06.E+00	7.93E-05
Step length left STD	-6.84.E-01	4.92E-01	-1.45.E+00	1.53E-01	7.81.E-01	4.36E-01
Cadence (steps/min)	7.66.E+00	8.99E-12	2.90.E+00	4.42E-03	5.47.E+00	2.66E-07
RL_balance Mean	-3.77.E+00	2.53E-04	-4.12.E+00	6.92E-05	7.40.E-01	4.60E-01
RL_balance STD	-3.06.E+00	2.48E-03	-3.38.E+00	1.07E-03	7.66.E-01	4.45E-01

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DISCLOSURE STATEMENT: The authors of this abstract have nothing to disclose.

Visual Field Distortion's Effect on Muscle Activation During Clinical Balance Assessment

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INTRODUCTION: Postural balance involves complex coordination and integration of multiple sensory, motor, and biomechanical components. Impairment/distortion of the visual field can affect the ability of the body to maintain balance. Increased influences on balance can result in activation of the motor cortex to enhance the responsiveness of the muscles in the lower body. **CLINICAL SIGNIFICANCE:** The purpose of this study was to examine the effects of visual field distortion on muscle activation in the lower extremity during a clinical balance assessment (CTSIB-M). **METHODS:** Twenty-one college-aged individuals (age 21.2 ± 2.4 yrs.) completed a stationary balance assessment with three different visual interventions (eyes-open, eyes-closed, and visual impairment). The balance assessment was conducted on a firm surface with each testing period lasting a duration of 30 secs. Surface electromyography was collected throughout each of the testing periods on muscle activation for the soleus, tibialis anterior, and medial gastrocnemius. **RESULTS:** Analysis revealed a significant increase in sway index for the eyes-closed condition ($p<.001$) and visual impairment condition ($p=.012$) as compared to the eyes-open condition. Also, a significant difference was reported between the eyes-closed and visual impairment conditions ($p=.038$). The eyes-open, eyes-closed, and composite scores for sway index were all significantly different from the age-matched norms ($p<.001$). No statistical differences were reported between the muscle activation for any of the muscle groups across the three conditions. **DISCUSSION:** These observed differences in muscle activation and sway index factors suggest that visual restrictions on interference affects stationary balance but does not directly correlate to an increase in muscle activation. **ACKNOWLEDGEMENTS:** West Chester University of Pennsylvania and the Department of Kinesiology. **DISCLOSURE STATEMENT:** No financial or incentive-based compensation was given or received for this study.

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Gait changes in young onset Parkinson's disease before and after medication

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PATIENT HISTORY A 48 year old male patient with young onset Parkinson's disease (YOPD) was referred for gait analysis. His main impairments consisted of lower extremity cramping, right upper extremity tremor, and right lower extremity weakness. The patient stated that his right leg "feels like cement". He was diagnosed with YOPD approximately 1.5 years prior to this analysis, suffering symptoms for approximately three years total. The patient reported sub-optimal management of motor symptoms with medication**.**

CLINICAL DATA The patient arrived in the afternoon for testing without taking his morning medication. Prior to data collection, a set of 41 retro-reflective markers were placed with respect to bony landmarks on the trunk, pelvis, thighs, shanks, and feet using a modified Helen-Hayes marker set. Two additional markers were located bilaterally on the medial femoral condyles and medial malleoli to define joint centers of the knee and ankle, respectively. One gait trial was collected prior to the patient orally administering his medication, Rytary. Gait data was subsequently collected every five minutes thereafter beginning at 30 minutes post medication administration, until 65 minutes post medication. Because the medication was administered orally, data collection began with a thirty minute delay, as no medication effect was expected before 30 minutes and over-fatigue was to be avoided. One additional trial was collected at 80 minutes post medication administration to insure maximal medicinal effect was captured.

MOTION DATA At 65 minutes post-medication, the patient demonstrated increased walking velocity, longer steps, and a narrower base of support (Fig. 1). The patient's single leg support increased bilaterally, demonstrating improved support capability for both lower extremities. Over the period of testing, temporal distance (TD) improvements were noted at the 30-35 minute interval with poorer performance after that time. TD factors then improved further during subsequent trials, corresponding to the pharmacodynamics of the immediate-release/extended-release characteristics of the patient's medication. At 65 minutes, the patient subjectively noted maximal improvement. The objective TD data supports the patient's subjective report. At that time, the patient's gait was the most symmetrical, with maximal improvements in most TD factors. At the final collection (80 minutes post-medication), TD factor improvement had leveled off in all factors except step width which did show further improvement.

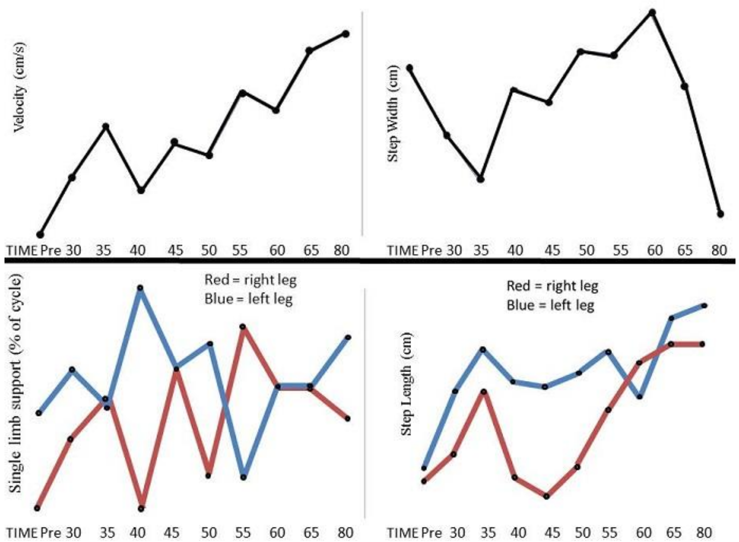

Figure 1: Changes in temporodistance factors over time after administration of immediate-release/extended-release medication. The data shows a positive effect at 35 minutes due to the immediate release and a subsequent improvement at 65 minutes due to the extended-release aspects of the medication.

The TD improvements are a result of subtle improvements in kinematics and kinetics. The more involved right lower extremity demonstrated greater kinematic improvements than the left lower extremity including improved loading response in stance and increased swing-phase flexion at both the hip and the knee. The right ankle reached neutral at initial contact to achieve heel strike when the medication was effective; while un-medicated the right ankle was plantar-flexed, striking at the midfoot. Overall, the kinematics demonstrated improved muscular control of the limb and foot, greater on the more involved right side, when medication was effective. As with the kinematics, moments tended toward normal when on-medication compared to off-medication, with greater improvement noted on the more-involved right side. Similarly the right side powers were closer to normal on-medication; while very little change was noted on the left side.

TREATMENT DECISIONS AND INDICATIONS Gait analysis improved clinical care for this patient by objectifying the patient's gait, providing the physician with important information on medication efficacy, and affording opportunities for patient education regarding treatment of YOPD.

OUTCOME Gait data was objectively recorded to improve clinical care of a patient with YOPD. Most notable improvements on medication were in TD factors including velocity, step width, single support time, and bilateral step length.

SUMMARY This case report is the first to quantify changes in the gait of a patient with YOPD off- versus on-medication. Gait analysis can provide practitioners with objective analysis of medication efficacy, and provide opportunities for patient education on treatment of YOPD.

DISCLOSURE STATEMENT There are no relevant financial relationships or off label usage to disclose.

EFFECTS ON BALANCE AND ON COGNITIVE TASK IMPACT WHEN VARYING FOAM THICKNESS BENEATH A RIGID PASSIVELY UNSTABLE SURFACE

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INTRODUCTION

A passively unstable surface (PUS) has previously been defined as a non-fixed surface whose movements are dynamically linked to an individual's motion responses[1]. For a rigid PUS, the platform on which an individual stands is inflexible, such that all portions of the standing surface experience the same angular motion. Foam, placed beneath a rigid PUS such as a wobble board, provides a restorative mechanism that imparts loads tending to return the PUS to an equilibrium position. Such a restorative mechanism can assist an individual with keeping a rigid PUS still, when standing on it.

In previous efforts at this institution[2], subjects instructed to keep a wobble board as still as possible demonstrated substantially improved balance performance when 5 inch thick foam was placed beneath the device. Substantial reduction of rigid PUS balance performance metrics (board tilt angle standard deviation and board angular velocity mean) was indicative of the balance task being considerably easier with foam beneath the wobble board. Such findings suggested that foam supported wobble board balance tasks might be achievable for persons unable to complete wobble board balance tasks without modification.

Earlier work, involving the single 5 inch foam thickness, was unable, however, to assess whether rigid PUS balance task difficulty could be decreased with lesser foam thicknesses. It was also not determined whether smaller foam thickness could trigger shifting, toward central body and upper extremity balance strategies, previously observed with a single foam thickness. The current study investigated the effects, on balance performance and mechanisms, of varying foam thickness beneath a rigid PUS, as well as foam thickness variation effects on how imposing a cognitive task impacts Rigid PUS balance performance and mechanisms.

CLINICAL SIGNIFICANCE

Understanding how supporting foam thickness mitigates difficulty of rigid PUS balance can assist with selecting a thickness to challenge an individual's balance capacity at a desired level. Such information can also clarify the limits to which rigid PUS balance difficulty can be abated.

METHODS

Ten healthy subjects (5 m, 5 f; 18-25 years) participated, after providing informed consent. For all trials, subjects were directed to keep a wobble board as still as able for 45 seconds. Trajectories for markers, placed in a Helen Hayes arrangement, were recorded (Qualisys) at 100 Hz. Subject performed 2 trials with each permutation of supporting foam thickness (0, 1, 2, 3, 4 & 5 inches), and concurrent cognitive tasks (no cognitive task, NC; verbal fluency task, VF). Performance metrics (based on task instruction) were standard deviations of board tilt angles (SD_{θ_bd}) and mean board angular velocities ($Mean_{\omega\_bd}$). Secondary measures were mean angular velocities of pelvis, trunk, upper arms, and lower arms ($Mean_{\omega\_pel}$, $Mean_{\omega\_tr}$, $Mean_{\omega\_ua}$, $Mean_{\omega\_la}$).

RESULTS

With respect to foam thickness, SD_{θ_bd} and $Mean_{\omega\_bd}$ increased significantly for each increment from 0 to 3 in., but did not differ from 3 to 5 in.; while with respect to cognitive task, these performance metrics were significantly greater with VF (Figure 1). With respect to foam thickness, Mean_{ω_pel}, Mean_{ω_tr}, Mean_{ω_ua}, and Mean_{ω_la} increased significantly for the increments between 0 and 2 in., but did not differ from 2 to 5 in.; while with respect to cognitive task, these secondary central body and upper extremity metrics were significantly greater with VF (Figure 2). There were no significant thickness/cognitive task interactions for any of the balance performance or secondary central body and upper extremity metrics.

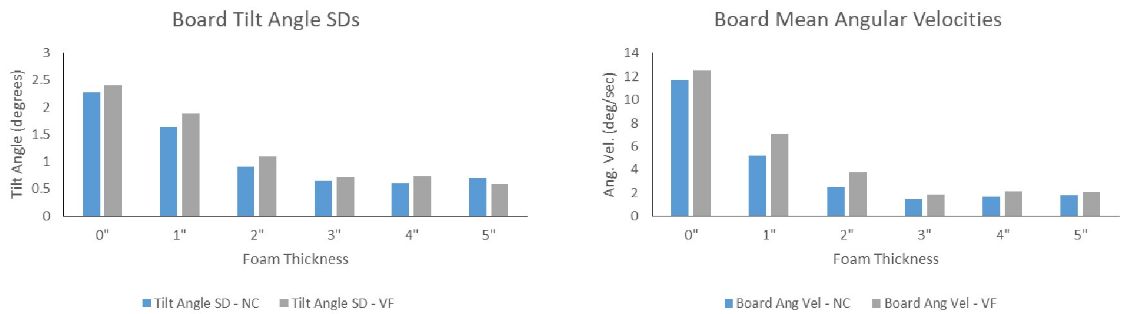

Figure 1. Balance performance measures for wobble board balance task.

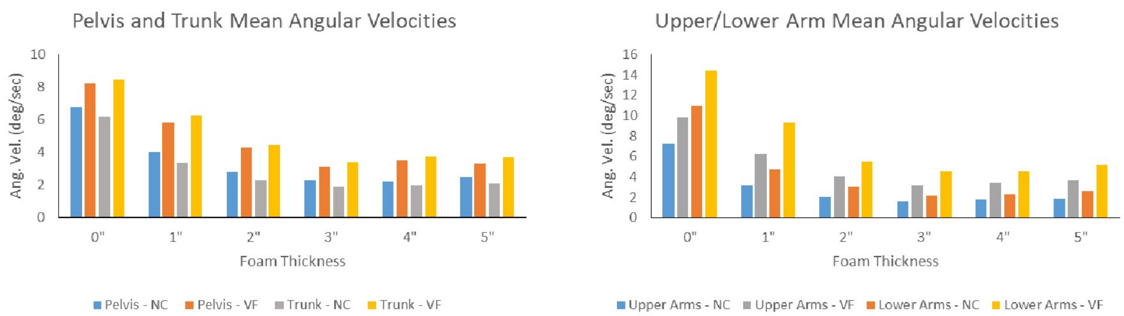

Figure 2. Central body and upper extremity angular velocities for wobble board balance task.

DISCUSSION

Task difficulty abatement with foam support (as indicated by improved performance metrics) was consistent with previous outcomes. However, current work further demonstrated that wobble board motion reduction, associated with 5 inch foam thickness, could be achieved with 2 inch less thickness. Task difficulty mitigation may enable more individuals to successfully perform rigid PUS balance tasks, however, these findings also suggest a limit, beyond which additional thickness increases do not further simplify these tasks. Central body and upper extremity motion reductions with foam were also consistent with earlier work; however, comparable lessening of central body and upper extremity strategies was attainable with 3 fewer inches of foam thickness. The consistency of cognitive task effects and the lack of significant foam thickness/cognitive task interactions was also of particular interest. Such results indicate that rigid PUS task difficulty reduction will not interfere with identifying potential impacts of concurrent cognitive tasks.

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DISCLOSURE STATEMENT

The authors have no conflicts of interest to disclose.

EXCESSIVE FOOT MOBILITY ENHANCES STATIC STABILITY UNDER VISUAL PERTURBATION

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INTRODUCTION

Balance and postural stability deficits have been linked to injury prevalence. Using a force plate to measure stability, Tropp et al. found male soccer players with increased postural sway to be at greater risk for ankle injury[1]. Plisky et al. found that even a simple balance test evaluating dynamic stability, such as the Star Excursion Balance Test (SEBT), was a reliable and predictive measure of lower extremity injury in high school basketball players[2]. Balancing on a soft surface has been linked to injury risk, suggesting that impaired proprioception may be a prominent influence on balance deficit[3]. Although poor balance and increased postural sway are recognized as predictors of injury, temporal and spatial aspects of instability may also be linked to foot type or foot mobility differences. Cobb et al. associated increased arch height with decreased medio-lateral postural stability on Time-To-Boundary (TTB) measurements[4]. Similarly, Hertel et al. reported greater mean Center of Pressure (COP) excursion in participants with high-arch compared to pes rectus foot postures during eyes-open single-limb–stance position testing[5]. COP displacement has been found to be a factor in ankle instability during the early stage of weight acceptance during lateral movement, potentially reducing capacity to dissipate ground forces upon impact and COP control is particularly challenged when the visual field is distorted. Linking foot type to instability, and instability to injury risk can offer insight into lower extremity injury mechanisms.

CLINICAL SIGNIFICANCE

Since static instability may be associated with lower extremity injury and foot mobility may be associated with instability, the purpose of this study was to examine the influence of foot mobility differences and visual perturbation on center of pressure displacement, measured by a sway path linear mean technique on a force plate.

METHODS

We tested 58 recreationally active, healthy male and female college age subjects (age = 21.12 ± 1.21 years, height = 66.67 ± 3.33 in., weight = 152.43 ± 23.92 lbs.). using a cross-sectional design and compared static stability by foot mobility levels both unperturbed and then under visual perturbation. The independent variables were foot mobility established by arch height index and navicular drop test to categorize the subjects into rigid or mobile foot type, and visual condition distorted with Fatal Vision goggles. The dependent variable was sway path linear mean, established as the average distance traveled between sample intervals collected over multiple 20 sec. one-legged static balance trials.

RESULTS

Independent t-tests revealed that a mobile foot group showed significantly lower sway path linear mean ($t = 2.05$, $p = .048$), compared to a less mobile (rigid) foot mobility type. However, these between group differences emerged only in the visually distorted condition, where higher foot mobility was associated with greater stability. Interestingly, static stability was not statistically different between foot mobility levels in the visual (eyes open) condition during balance testing.

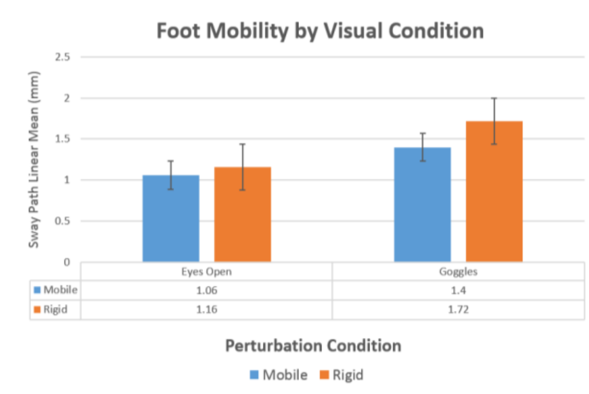

Figure 1. Stability: foot mobility by visual condition.

DISCUSSION

Since group differences occurred only in the visually perturbed condition, we infer that foot mobility appears to affect static stability when appreciation of visual field is diminished and subjects are left to rely on other substrates of postural control for balance. Lower leg strength was not assessed. However, extrinsic muscles acting on the foot and utilized for misalignment correction may vary in their capacity to affect stability. Based on potential , but untested, strength differences, it is possible that some foot types effectively conform (in functional malleability) to loading stress while others (presumably weaker subjects) might succumb to loading stress and this may, in turn, permit the display of instability potentially associated with injury risk. Future researchers should investigate the effect of functional strength differences on static and dynamic stability across various perturbation conditions.

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ACKNOWLEDGMENTS

DISCLOSURE STATEMENT

Authors have no conflicts of interest to disclose.

BALANCE CONTROL DURING SQUAT AND LUNGE EXERCISES IN OBESE INDIVIDUALS

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INTRODUCTION

Obesity rates have increased significantly in the United States over the years, with approximately 30% of adults now considered obese. It has been demonstrated that in static stance, obese individuals have poorer postural control than normal weight individuals [1]. Very limited number of studies have looked at analysis of balance during common functional strengthening activities. The squat and lunge are both commonly utilized activities in physical therapy, and may challenge balance control of obese patients during these activities. The purpose of this study was to analyze the balance of obese females while performing squat and lunge activities, as measured by center of pressure area (CoP area) and center of pressure velocity (RMS-V).

CLINICAL SIGNIFICANCE

Clinicians may be able to better tailor exercise programs to this patient population, which could increase performance and decrease the risk for loss of balance, falls, and injury.

METHODS

Twenty females (10 obese and 10 normal weight) were analyzed for CoP area and RMS-V during standing, squat and lunge activities at three difficulty levels (squat: 60°, 70°, and 80° knee angle; lunge: step of 1.0, 1.1, and 1.2 times tibial length).

Kinematic data were collected using an Optotrak motion analysis system (Model 3020, Northern Digital Inc., Waterloo, Ontario, Canada) operating at 60 Hz, and were filtered at 6 Hz, using a zero phase lag, fourth-order, Butterworth low pass filter. Kinetic and COP data was collected by using Kistler force plates (Kistler, NY).

Visual 3D software (C-Motion Inc. Kingston, Ontario) was used to perform link-segment calculations. The values of CoP area and RMS-V were calculated for the individual trials of each difficulty level of the lunge and squat. For the standing trials, 30 seconds of sway data were collected. The entire 30 seconds were analyzed as a whole, and seconds 2-5 were analyzed to mimic the squat and lunge data collected (3 seconds). A repeated measures ANOVA was performed to analyze between- and within- group difference. Between-group effect size was also calculated as an indication of clinical relevance.

RESULTS

A total of 20 subjects were recruited for the study, 10 obese females with a mean BMI of 39.23 ± 6.67 kg/m^2 and mean age of 40.2 ± 3.0 years and 10 age matched normal-weight females (BMI 18.5-24.9 kg/m^2) with a mean BMI of 21.50 ± 1.55 kg/m^2

For standing with eyes open, 30 seconds, there was a significant difference between obese and normal weight females for RMS-V (p=0.011), but not for CoP area (p=0.070). For squatting, changing the depth of the motion did not result in a significant change in CoP area (p=0.401) or RMS-V (p=0.057). Altering the step distance for lunging did not result in any change in CoP area (p=0.297) or RMS-V (p=0.412). There was a significant difference between obese and normal weight females for CoP area (p<0.001) and for RMS-V (p=0.005) during the lunge activity.

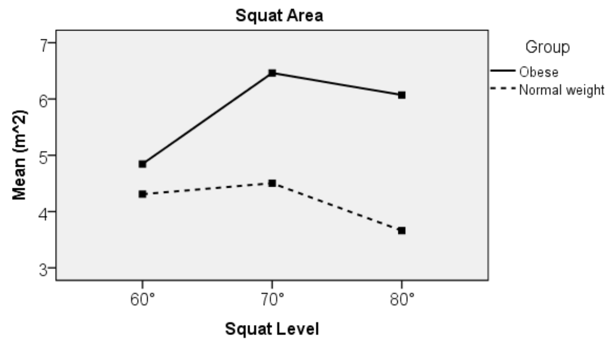

Figure 1: Shows the squat area for 10 obese and 10 normal weight subjects performing squat at three different levels (60, 70 and 80 degrees of knee flexion).

Table 1: Represents COP area and RMS-V for standing, squat and lunge activities.

	30 sec stand	3 sec stand	Squat			Lunge		
			60°	70°	80°	x1	x1.1	x1.2
COP Area								
BMI	0.56	0.41	0.13	0.16	0.21	0.65	0.77	0.70
Waist circumference	0.57	0.41	0.08	0.23	0.11	0.56	0.70	0.70
RMS-V								
BMI	0.70	0.57	0.17	0.22	0.43	0.45	0.69	0.64
Waist circumference	0.71	0.55	0.06	0.24	0.43	0.34	0.59	0.64

DISCUSSION

During lunging activities, obese females demonstrated poorer balance control than their normal weight counterparts. This difference was not seen during the squatting activity, though a trend was revealed with a medium effect size. Because of this finding, clinicians may want to consider balance control as a variable when prescribing lunge activities to obese individuals.

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DISCLOSURE STATEMENT

Bhupinder Singh or any co-authors have no conflicts of interest to disclose.

INDICATORS OF FSHD QUANTIFIED AS DECLINES IN SPATIO-TEMPORAL GAIT CHANGES DURING SINGLE AND DUAL-TASK WALKING

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INTRODUCTION

Facioscapulohumeral muscular dystrophy (FSHD) is a common form of muscular dystrophy that causes muscle wasting and weakness [1,3]. The progressive muscular atrophy specifically in the lower extremity affects postural control which influences the ability to navigate one's environment successfully. The lower extremity weakness increases the risk of falls specifically when secondary tasks are added while performing any primary functional tasks such as walking [2].

CLINICAL SIGNIFICANCE

Currently, there is no cure for FSHD and there is a dire need to create ecologically valid assessments that can show the disease progression in functional daily tasks.

PURPOSE

The purpose of the current study was to quantify the changes in spatio-temporal gait characteristics during normal (single task-ST) and dual-task (DT) walking on a Strideway gait pressure mat system.

METHODS

Eight FSHD (M±SD = 52.75±15.71 years) and eight nearly matched healthy controls (M±SD = 51±17 years) were asked to perform five ST and five DT walking in a pseudorandomized order. The DT included a cognitive serial 7's subtraction task from a random number between 50-100. Gait dependent variables: cadence (steps/min), gait time (sec) and gait velocity (cm/sec) were obtained from Tekscan Strideway (7.7X, sampling frequency 30Hz, 10 ft mat).

RESULTS

Two specific approaches: pairwise comparison, where each FSHD was matched with their control and Multilevel modelling (MLM) approach, to explain the group and task effects for all variables were taken. The pairwise comparison showed that Cadence was significantly different for both ST ($p<0.0078$) and DT ($p<0.039$) where FSHD showed lower cadence compared to controls. Gait time was significantly higher for both ST ($p<0.0078$) and DT ($p<0.0078$) in FSHD compared to controls. Lastly, gait velocity was significantly lower for FSHD during ST ($p<0.0078$) and DT ($p<0.016$). MLM approach revealed a Group x Task interaction for cadence (F (1, 191.04) = 11.782, $p < 0.001$) and Gait velocity (F (1, 191.04) = 11.782, $p < 0.001$). Gait time only had a Group and Task main effect where FSHD took more time overall compared to controls to complete ST and DT. However, FSHD did not show a dual task cost effect when compared to ST for gait time. All means (SDs) are provided in table 1 for all variables.

TABLE 1: Mean (SD) of all dependent variables of interest. P-values are reported for the omnibus RMANOVA.

	Control		FSHD		
Variables	ST	DT	ST	DT	p-value
Cadence	104.677 (13.589)	84.939 (20.775)	80.105 (14.015)	68.833 (13.700)	0.001***
Gait Time	2.242 (0.516)	3.248 (1.242)	4.103 (1.562)	5.362 (2.346)	0.001***
Gait Velocity	113.782 (22.073)	89.039 (29.779)	72.328 (26.944)	60.115 (24.300)	0.001***

Note. ST = Single task; DT = Dual task; * indicates p<0.05, ** p<0.01, **p<0.001

CONCLUSION

Our findings indicate specific pairwise differences in both ST and DT walking observed in the gait parameters as decreased cadence and gait velocity and increased gait time during ST and DT in FSHD compared to controls (Table 1). Additionally, the MLM results showed that controls exhibited the dual-task cost as expected but FSHD did not across all variables. In conclusion, individuals with muscular dystrophy may not exhibit dual-task costs due to the dominance of the affected primary walking task as compared to healthy controls indicating decline in lower extremity muscles. Further research is needed to explore changes in such costs over multiple time points to express disease progression in normal walking.

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DISCLOSURE STATEMENT

Sushma Alphonsa is a Research Scientist in the Neuromechanics Laboratory, School of Community Health Sciences at UNR. This project was a collaboration with the Department of Pharmacology, UNR School of Medicine. All authors on the project have no conflicts of interest to disclose.

Comparing short-term outcomes of selective dorsal rhizotomy between conus medullaris and cauda equina techniques

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INTRODUCTION: Selective dorsal rhizotomy (SDR) is performed via a single-level technique at the conus medullaris (conus) or a multi-level technique at the cauda equina (cauda).[1,2] Few studies have assessed differences in outcomes between the two.[3,4] The aim of this study was to determine whether these two surgical methods lead to equal short-term outcomes (1-3 years post-SDR) for individuals with spastic cerebral palsy.

CLINICAL SIGNIFICANCE: Our investigation can be used to help guide treatment decision making for clinicians and families.

METHODS: This was a retrospective chart review. Participants met the following criteria: diagnosed with cerebral palsy, underwent SDR at our institution between 2013 and 2017, were <11 years old at time of SDR, had a pre-SDR gait analysis ≤18 months of surgery, had a post-SDR gait analysis 8-36 months after surgery, and had no major surgery between pre- and post-SDR visits other than the SDR. The choice of SDR technique depended on surgeon and family preference. Between and within-group data were compared using the Chi-square or t-test, as appropriate.

RESULTS: Among 80 children included in this study, 21 and 59 children underwent conus and cauda SDR, respectively. Pre-SDR demographics, functional measures, and physical examination (PE) were well-matched. Gait characteristics were matched at the knee; however, there were baseline differences in overall gait pathology, particularly at the ankle.

Gait kinematics: During gait, pre-post sagittal plane knee kinematics improved for both groups at initial contact (conus = +11°, cauda = +11° more extended; p<0.01), in stance phase (conus = +9°, cauda = +10° more extended; p<0.01), as well as in swing phase (conus = +4°, conus = +4° more flexed; p<0.01). However, improvements in sagittal plane ankle kinematics differed significantly between techniques post-SDR (Figure 1). Between pre and post-SDR, mean increases in stance phase and swing phase dorsiflexion were significantly larger for the cauda group compared to the conus group (conus: +2° in stance, +0° in swing; cauda: +10° in stance, +7° in swing).

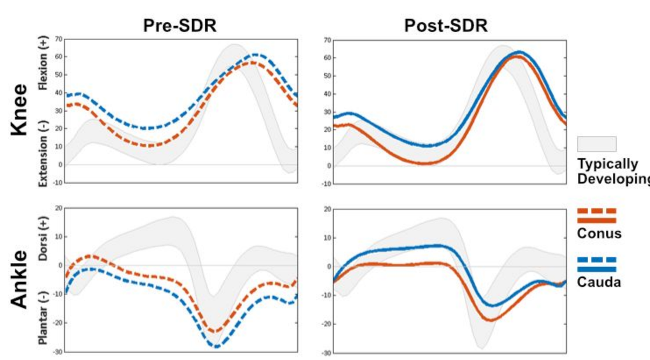

***Figure 1**. Knee and ankle kinematics show residual equinus in the conus group.*

Gait Deviation Index (GDI):[5] The cauda group had lower pre-SDR GDI (conus = 68.8±11.4, cauda = 63.5±10.5; $p<0.01$; Figure 2). Post-SDR GDI of both groups were similar (conus = 69.0±13.7, cauda = 67.8±10.3; $p=0.55$). The pre-post change in GDI was greater for the cauda group (conus = +0.2, cauda = +4.2; $p=0.05$).

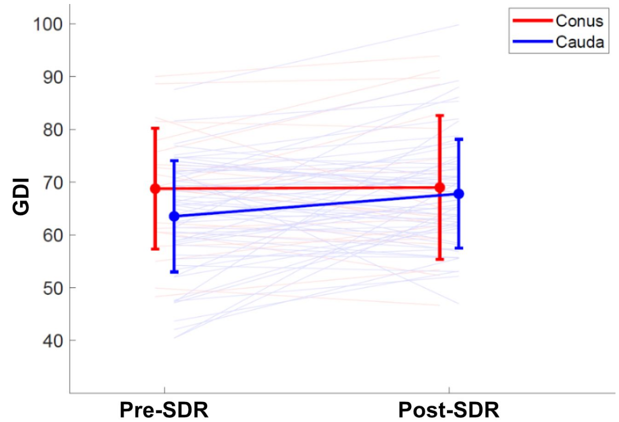

Figure 2. *Pre-SDR and post-SDR changes of GDI in conus and cauda groups.*

Physical examination: Most PE measures (*e.g.*, strength, spasticity, joint range of motion [ROM]) exhibited a similar improved response. Among these, passive ROM of the ankle (knee flexed and extended) improved equally for both groups (conus = +3-5°, cauda = +4-5° more dorsiflexed; $p<0.01$), as did the popliteal angle (conus = +14°, cauda = +11°; $p<0.01$). Post-SDR static spasticity, as measured by Ashworth, was normalized for both techniques.

Gross Motor Function Measure (GMFM):[6] The 66-item GMFM (GMFM-66) was collected on a subset of participants due to variation in our institution's clinical routine (conus, n = 7; cauda, n = 16). The conus group had higher scores compared to the cauda group at both pre-SDR (conus = 68.4±8.8, cauda = 61.1±8.9; $p=0.09$) and post-SDR (conus = 74.1±11.0, cauda = 64.8±8.7; $p=0.04$). Pre-post changes in GMFM-66 scores improved for both groups (conus = +5.7 [$p= 0.04$], cauda = +3.7 [$p<0.01$]).

DISCUSSION: Conus and cauda SDR techniques result in similar short-term outcomes for spasticity, gross motor function, and passive joint ROM. Additionally, pelvis, hip, and knee kinematics in all three planes were largely equivalent at post-SDR. However, a significantly different response in sagittal plane ankle kinematics was seen. Post-SDR dorsiflexion improved more in the cauda group resulting in a more normalized stance phase motion and position. In contrast, the changes observed at the ankle in the conus group were relatively small, resulting in residual equinus. Interestingly, static physical exam measures, which were well-matched pre- and post-SDR, failed to explain these clear dynamic gait differences. The causes and implications of these differences merit further investigation.

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Hemiplegic Gait: How good is the good leg?

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Purpose:
The gait alterations seen in hemiplegic cerebral palsy are not well understood. Most studies looking at gait patterns in hemiplegic cerebral palsy only look at the more affected side, and very few examine the contralateral side.[1-4] The purpose of this study was to analyze the findings of the contralateral (stronger) limb using physical examination, kinematics, kinetics, electromyography, and pedobarography along with brain imaging to determine whether any specific patterns could be identified that could aid in clinical diagnosis and post-operative recovery.

Methods:
Retrospective review of all children referred to our gait lab with a diagnosis of hemiplegic cerebral palsy from August 2018 through June 2019. Age, brain imaging, cause/timing of injury, family/child goals and concerns, 3D motion analysis with gait kinematics, kinetics, electromyography, pedobarography, with PODCI and CP CHILD outcome measures, and radiographs were reviewed. Gait study recommendations and family's chosen treatment were reviewed. Post-surgical 3D motion analysis, PODCI and CP CHILD results for those greater than 6 months from selective dorsal rhizotomy and greater than 1 year from orthopaedic interventions were reviewed.

Results:
Eighteen patients were referred with a diagnosis of hemiplegic cerebral palsy. Of the 18, 16 had brain imaging for review. Of the 16 that had imaging, 12 had findings in the ipsilateral hemisphere.

Of the children who had only contralateral findings on MRI (4/16), several patterns emerged on their stronger leg: 100% increased femoral anteversion and 75% external tibial torsion, 75% GMFCS I, 75% limb-length discrepancy, 75% prolonged activity of the hamstrings, 75% Gait Profile Score (GPS) of <6 on their "unaffected" leg, 50% decreased knee extension in terminal swing, and 50% ankle plantarflexion after initial ankle dorsiflexion. Children with only ipsilateral findings on brain MRI and "hemiplegia", 50% underwent SDR, 25% SEMLS, and 25% non-operative interventions to date.

Of those with a diagnosis of hemiplegic cerebral palsy and ipsilateral findings on MRI (12), the following were observed on their stronger leg: 92% increased femoral and/or tibial torsion on exam and kinematics, 92% GPS >6 on their "unaffected" leg, 83% evidence of gastrocnemius spasticity on physical exam, decreased heel contact on the less affected side, and early ankle plantarflexion after initial contact, 83% decreased knee extension in terminal swing, 83% leg length discrepancy with a stronger side that was longer, 50% co-contraction of the quadriceps/hamstrings and/or tibialis anterior/gastrocnemius, 25% decreased peak knee flexion in swing, 17% short and slow hamstrings on modeling, 8% evidence of hamstring and psoas spasticity on sagittal kinematics (double bump pelvis), and no posterior pelvic tilt in any cases. Children with global findings on brain MRI and “hemiplegia 83% have undergone interventions to date: 42% non-operative, 33% orthopaedic interventions (50% bilateral, 50% unilateral), and 25% selective dorsal rhizotomy.

Conclusion:
Beware that in children with a diagnosis of hemiplegic cerebral palsy, the stronger leg is also affected by abnormal gait, tone, and deformity. Those with global findings on brain MRI, or findings on the ipsilateral side of the perceived hemiplegic side, will have gait deviations on their stronger side.

Significance:
The identification of pathologic, non-compensatory findings on the “good” leg can help guide interventions to improve gait kinematics on the contralateral side and in pre-operative counseling for families. This may decrease energy expenditure and improve comfort in this patient population by thorough evaluation of the stronger side in children with “hemiplegic” cerebral palsy. When considering surgery for an individual with “hemiplegic” cerebral palsy, recognize that the “good” leg may not be as good as one thinks and may cause issues with weight bearing and ambulation post-operatively.

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A Simple Neural Reward Circuit May Motivate Human Gait Development and Even Explain Cerebral Palsy Gaits

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INTRODUCTION

The assumption that the human bipedal gait emerges from an innate pattern encoded in the motor control system may be incorrect. Instead, our gait may be explained by assuming a mutation that created, in a human ancestor, a simple neural reward circuit. We conjecture a circuit exists that rewards the sensation of Rhythmic Audible Vestibular Jolts (RAVJ). Because rewards motivate the behaviors that can produce them (known as operant conditioning), a RAVJ reward will motivate RAVJ producing behaviors. One of those behaviors is learning our bipedal gait since human gaits (from first steps to adult style and even cerebral palsy gaits) do generate RAVJ. This occurs because the foot to floor impact creates a compression shockwave in a foot bone that travels via the skeleton to the head. At the head, that shockwave produces a strong vestibular jolt and it is audible (as a click or boom from the shockwave fluid mechanics in the cochlea as occurs in bone conduction hearing). The periodic nature of our foot-strikes thus generates RAVJ. Additionally, because successfully avoiding an aversive outcome also acts as a reward [1], and falling can be painful, gait modifications that reduce the aversive sensations of falls will be rewarded and thus learned [2]. Does the gait literature support this novel RAVJ reward conjecture?

CLINICAL SIGNIFICANCE

If the RAVJ reward conjecture is correct, then cerebral palsy gaits (such as toe strike and severe crouch) would occur because of the CP inability to develop the walking reflex that precisely stops forward motion of the foot prior to floor contact. Without that reflex, painful slips will occur more often [3]. Thus the CP patient, in order to prevent slips, will discover the foot strikes that reliably stop forward foot motion (the toe and crouch strikes); however, because the strike must still generate a strong enough shockwave to the head, the strikes are high impact and thus lead to skeletal damage. This new gait theory opens new therapy options to prevent developing the damaging CP gaits and/or to prevent skeletal damage.

METHODS

Literature search and review of studies on gait with data predicted by the RAVJ reward theory. Studies included: infant first steps; gait development and biomechanics; cerebral palsy gaits and development; slip and trip fall mechanics; reflex plasticity; and foot impact mechanics. Additional studies are from trained monkey bipedal gaits and from hominine evolution (Ardi).

RESULTS

The RAVJ reward conjecture predicts human gaits (in the absence of painful steps) will normally produce RAVJ; it also predicts gait modifications are discovered by experience that reduce falls (and quickly stop the associated fall sensations) but that still produce an adequate shockwave. All data found was consistent with the conjecture's predictions (see discussion).

Also, since monkey training shows targeted rewards can produce upright bipedalism and reflexes without needing neural specializations [4], then neither do humans. Also, the origin of the RAVJ reward neural circuit would have made our style of bipedalism immediate in some ancestor; this is consistent with the *Ardipithecus* interpretation [5].

DISCUSSION

An infant's first steps are hard collisions of their feet to the floor (shockwave producing) [6] and is consistent with a RAVJ reward motivating first steps. Infants then quickly learn to prevent painful slip falls by stopping the foot's forward motion prior to floor contact and are developed into reflexes [7]. Because a trip fall can also be painful, all falling sensations also become aversive by association; therefore, new gait modifications are adopted which maximize trip recovery time. Maximizing recovery time is accomplished by keeping the body COM (HAT) behind a foot (either stance or the swinging foot) for as much distance and time as possible during the leg's swing phase; however, those gait modifications, while increasing trip recovery time, must still produce an adequate shockwave to the head. Thus, the new walker's first learned modification (within 1 year) is a straight knee (which maximizes shockwave transmission) and with the heel down (which minimizes energy absorption by the heel pad). This gait modification is the *inverted pendulum* which allows a much lower velocity foot strike while still producing adequate RAVJ. This gait allows the COM to be further back from the foot strike area which provides more time for the swing foot to move ahead of the body before the body passes the stance foot; hence, it increases trip recovery time and is thus rewarding when recovering from trips. Discovery of additional gait modifications continues until, in the mature gait, the COM is always behind one of the feet [8] while still producing RAVJ. A part of that strategy is to reduce the HAT forward velocity until mid-swing (by the first rocker and knee flex-wave); another is to also propel the swing leg forward as quick as possible without accelerating the HAT [9]. Every adult gait element (even arm-swing) can be shown to maintain a more rearward HAT but still maintain an adequate vertical foot velocity at heel impact (needed to produce a strong enough shockwave) [10]; that more rearward HAT rewards by increasing the trip recovery time; therefore those gait elements are learned by experience. The simple and plausible RAVJ reward conjecture provides a strong explanatory and predictive mechanism for human gaits, including cerebral palsy gaits.

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ANTERIOR DISTAL FEMORAL HEMIEPIPHYSIODESIS WITH AND WITHOUT PATELLAR TENDON ADVANCEMENT FOR FIXED KNEE CONTRACURES IN CHILDREN WITH CEREBRAL PALSY

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INTRODUCTION:
Distal femoral extension osteotomies (DFEOs) are often combined with patellar tendon advancement to improve fixed knee flexion deformities with associated patella alta and crouch gait. Anterior distal femoral hemiepiphysiodesis (ADFH) has become an increasingly popular alternative to DFEO for patients with open physes but is usually done without concomitant patellar tendon advancement (PTA). The purpose of this study was to compare ADFH alone versus ADFH with PTA for the treatment of fixed knee flexion contractures and crouched gait in children with cerebral palsy (CP). We hypothesized that there would be larger and faster correction with combined ADFH and PTA.

CLINICAL SIGNFICANCE:
ADFH, alone or when combine with PTA, is an effective procedure for children with fixed knee flexion contractures. Combined surgery of ADFH and simultaneous PTA may be more effective for children with larger knee flexion contractures than ADFH alone.

METHODS:
Retrospective review was performed of pre- and postoperative gait analysis data and perioperative knee radiographs for children with CP with fixed knee flexion contractures who underwent ADFH alone, or ADFH with PTA. Data were analyzed using t-tests, chi-square and multiple regression analysis to control for covariates that differed between groups at baseline.

RESULTS:
Twenty-five participants (42 limbs) were included, 17 male, mean age at surgery 12.9 (SD 1.9 years). Both groups experienced significant improvement in popliteal angle, knee extension range of motion and knee extension in the stance phase of gait. (Table 1) Greater improvement was seen for all variables in the combined ADFH/PTA group. However, multivariate regression analysis indicates that the differences were due to greater contracture and knee flexion during gait preoperatively in the combined ADFH/PTA group ($p<0.02$), and not to the procedure performed ($p> 0.55$). Significant improvement was seen in all groups when stratified by presence or absence of radiographic evidence of patella alta, also related to severity of deformity rather than the procedure performed. Rate of contracture resolution ranged between 0.5 – 1.2° per month, faster in larger contractures ($p=0.007$).

Table 1: Within and between group comparisons of change in knee variables (degrees).

	ADFH (N= 25 limbs)			ADFH/PTA (N= 17 limbs)			*p-value* **Change between groups** b	
	Pre Mean (SD)	Post Mean (SD)	*p-value Pre- to post* [a]	Pre Mean (SD)	Post Mean (SD)	*p-value Pre- to post* [a]	*Univariate*	*Multivariate*
Popliteal angle	51 (13)	39 (13)	*<0.0001*	62 (10)	38 (9)	*<0.0001*	*0.002*	*0.41*
Knee extension ROM	-13 (5)	-5 (7)	*<0.0001*	-19 (8)	-4 (7)	*<0.0001*	*0.006*	*0.44*
Maximum knee extension stance	-31 (9)	-15 (18)	*0.0001*	-51 (16)	-17 (14)	*<0.0001*	*0.003*	*0.52*
Rate of change in knee extension ROM	0.52 (0.37)			1.02 (0.71)			*0.005*	*0.32*

ROM – range of motion
[a] p-value based on paired t-test
[b] P-value based on univariate regression and multivariate regression (controlling for preoperative popliteal angle, knee extension ROM and maximum knee extension in stance)

DISCUSSION:
ADFH, both with and without PTA, is effective in improving both passive and dynamic knee extension in skeletally immature patients with CP, correcting contractures at a rate of 0.5 to 1.2° per month. Combined ADFH and PTA surgery may be preferable to ADFH alone in patients with larger contractures of up to 30-35°. Additional research is needed to determine the upper limits of contractures which can be effectively treated with ADFH.

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Long and Short Term Kinematic Gait Outcomes following Rectus Femoris Transfers in Ambulatory Children with Cerebral Palsy

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PURPOSE

Rectus femoris transfer (RFT) is utilized to improve a stiff knee gait in ambulatory children with cerebral palsy (CP). There is limited data regarding long-term outcomes of RFT. The purpose of this study was to assess long and short term kinematic gait outcomes after RFT in ambulatory children with CP.

METHODS

A retrospective review identified ambulatory children with spastic diplegic CP, who had RFT and had a motion analysis preoperatively and a 1 year post-operatively. We also identified those with 5 and 10 year post-operative motion analysis. Three primary kinematic variables were assessed: peak knee flexion range of motion in swing (PKFSW), timing of peak knee flexion in swing as percent of the gait cycle (PKF%GC), and total knee range of motion in swing (KROM). Since clinical interpretation of gait kinematics identifies how far from normal reference values a child is, the values for each outcome variable were calculated in standard deviations (SD) from the mean of the normal reference value for each variable. Responders (those who moved towards the mean) and non-responders (those who moved away from the mean) were identified. Mobility level, identified by the Gross Motor Functional Classification System (GMFCS) level, was evaluated as a predictor of response.

RESULTS

One hundred and twenty-one ambulatory children (237 limbs) with spastic diplegic CP who had RFT were included (mean age=10 years; GMFCS Level II= 67 limbs & III =52 limbs). All participants (237 limbs) had a preoperative and one year postoperative motion analysis. Motion analysis at 5 and 10 years post-operatively included 85 limbs and 28 limbs, respectively.

Eighty-four percent of children improved in both PKFSW and PKF%GC with a significant difference in mean preoperative values for responders and non-responders ($p < 0.001$). PKFSW improved in 59% of limbs. Responders started 1.4 SD below the mean PKFSW preoperatively, and improved by an average of 2.2 SD to reach a normal range at 1 year post-operatively. Improvement was maintained at 5 and 10 years postoperatively. Those at GMFCS level II were more likely (OR 1.3, CI 1.05, 1.5) to have improved PKFSW at 1 year postoperatively than those at GMFCS level III.

PKF%GC improved in 63% of limbs. Responders had delayed PKF%GC, starting 0.09 SD above the mean (later in the gait cycle) preoperatively. Their timing improved towards normal values: 0.05 SD, 0.06 SD and -0.01 SD (earlier in the gait cycle) at 1, 5 and 10 years postoperatively, respectively.

Although KROM improved in only 32% of limbs, there was again a significant difference in mean preoperative values between responders (-5.8) and non-responders (-3.5) ($p<0.001$).

CONCLUSIONS

RFT improves short and long-term kinematic gait outcomes. The majority of children responded to RFT with improvements in PKFSW and PKF%GC at 1, 5, and 10 years post RFT. GMFCS level is a predictor of improved PKFSW, with children at GMFCS Level II having an increased likelihood of improvement at 1 year post surgery. Characteristics associated with responders who maintain long term positive outcomes need to be identified. Optimizing surgical techniques with immediate postoperatve weight bearing offers many advantages with few disadvantages.

IMMEDIATE EARLY WEIGHT BEARING REHABILITATION FOLLOWING BONE PROCEDURES IN CEREBRAL PALSY

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INTRODUCTION

Single Event Multilevel Surgery literature reports wide variations in surgical practice and rehabilitation protocols. Most include 2-6 weeks of non weight bearing. Disadvantages of this include weakness, contractures, decreased independence, social isolation, depression and delayed rehabilitation.

CLINICAL SIGNIFICANCE

We propose that immediate postoperative weight bearing leads to improvements in functional outcomes, decreasing family burden of care and recovery time with minimal disadvantages. The study's purpose is to assess complications and outcomes with immediate postoperative weight bearing after bony surgery in ambulatory cerebral palsy patients.

METHODS

The study is a retrospective longitudinal cohort with 126 consecutive patients from 2004 to 2017 who underwent any bony lower extremity surgery with minimum 2 years postoperative followup treated by a single surgeon at a community hospital. All patients began standing on postoperative day 1 and were taking steps by day 2; those with foot procedures by day 3. Medical records and radiographs were reviewed to determine the frequency and type of complications graded according to the Modified Clavien Dindo (MCD) system; including medical, wound, infection, union, or hardware issues, and loss of graft position. FMS scores preoperatively and at 12 and 24 months postoperatively were also recorded.

RESULTS

Nineteen had unilateral and 107 bilateral cerebral palsy. 102 patients had bone procedures as part of SEMLS and 24 did not. GMFCS levels were as follows: I-4, II-80, III-36, IV-6. Fifty-Six had unilateral and 70 had bilateral surgery. Twelve patients had staged surgeries (17 days to 11 months apart). There were an average of 2.87 bone and 7.13 total procedures per patient. There were 2 iliac, 64 proximal femoral, 36 distal femoral extension, 19 tibial tubercle, 85 distal tibial, 83 calcaneal, 34 medial cuneiform, 4 first metatarsal, and 4 phalangeal osteotomies. There were 9 subtalar, 3 talo-navicular, 3 calcaneo-cuboid, 9 naviculo-medial cuneiform, 3 first metatarsal- medial cuneiform, and 3 first MTP arthrodeses. There were 20 Grade I/II MCD, 4 Grade III, and one grade IV and V MCD complications. Most patients' FMS scores were the same or improved compared to baseline at 2 years postoperatively.

DISCUSSION

Immediate weight bearing after bony surgery in ambulatory CP patients appears safe and effective. The complication rate is no higher than in historical cohorts. FMS scores did not appeared negatively effected by early weight bearing. Study limitations included a heterogeneous population, retrospective design, and no control group. Study strengths included a single surgeon with uniform postoperative protocols.

Optimizing surgical techniques with immediate postoperative weight bearing offers many advantages with few disadvantages.

Velocity-Matched Normative Comparison of Gait Patterns of Individuals with Chronic Traumatic Brain Injury

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INTRODUCTION

A proper framework for evaluating TBI gait data can clarify understanding of deficits and guide treatment decisions. Past research to describe and understand gait deficits in the TBI population has attempted to frame TBI gait data in such a way, but the reference data to allow this do not seem to exist. Walking speed has a significant influence on magnitude and timing of many gait parameters. Consequently, many prior evaluations of TBI walking data have been compared to available or convenience (i.e. not velocity matched) gait data.

CLINICAL SIGNIFICANCE

Lack of appropriate reference data can lead to misconceptions about TBI gait as well as unrealistic expectations or treatment goals.

METHODS

Data were obtained from a prior trial of twenty-three participants with waking limitations due to chronic TBI participated in a trial to determine the efficacy of a variety of gait training interventions on follow up gait performance [1]. The data presented here were prior to any interventions were administered. Participants were at least 18 years of age and had a traumatic brain injury (≥12 months). All subjects provided informed consent. Prior to enrollment, participants were able to ambulate at least 10 meters at a self-selected velocity between 0.2 m/s and 0.6 m/s. If needed, braces (ankle/foot, knee orthoses) or upper extremity assistive devices (canes, walkers) were allowed. Overall, one subject used a MAFO, and all but three used upper limb support (parallel bars) during walking trials. Subjects were instrumented with bilateral lower extremity and trunk markers and walked across the 20-meter laboratory walkway that included 5 embedded forceplates at their self-selected velocity. Subjects typically were able to take at least 2-3 steps prior to entering the calibrated recording volume. Subjects had to wait on the order of 5 seconds once reaching the end of the walkway and turning around – which minimized the discontinuity and contributed to the steady-state nature of the data collected.

RESULTS

Bilateral 3D joint kinematics and kinetics were computed. Sagittal findings are reported here – a sample of which are shown in the below figures.

DISCUSSION

This is the first report, to our knowledge, of a systematic data collection of gait data from individuals with chronic TBI along with comparison to velocity matched reference normal data. Data were collected using the same system, same staff and processed using the same biomechanical model. Our data support that there is, as has been speculated, increased variability in TBI gait. Nearly all of the *basic patterns* observed in healthy normal gait were present in the

TBI group – a finding that could not be made unless comparisons to appropriate velocity-matched data. Timings and magnitudes of specific phases were altered and these can have important consequences on subsequent interventions and strategies to improve walking. We believe these data can serve as an important tool for the clinicians as they evaluate and develop treatment plans for this population.

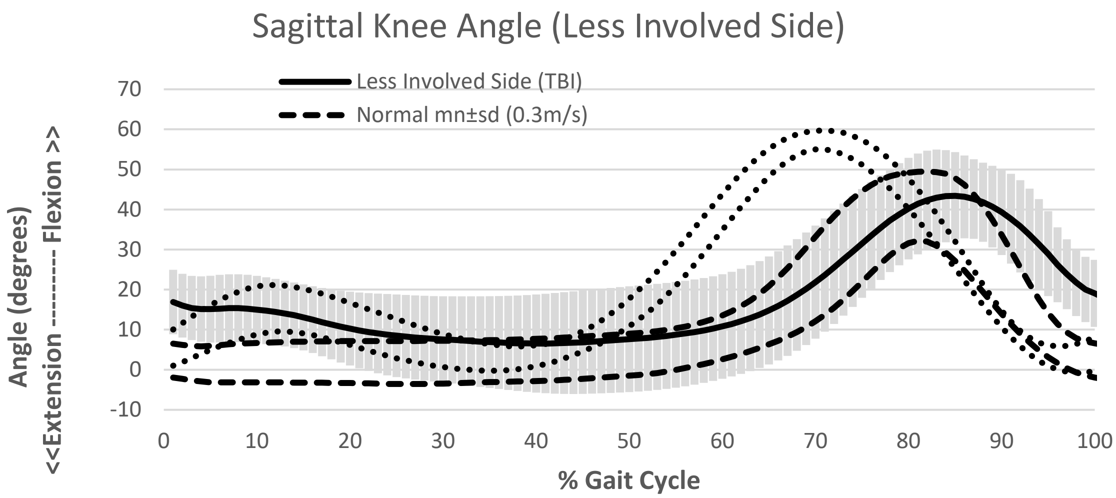

Figure 1. Sagittal Knee Angle Data for Both TBI and Normal Groups Walking at ~0.3m/s along with Healthy Normal Subjects Walking at 1.2m/s.

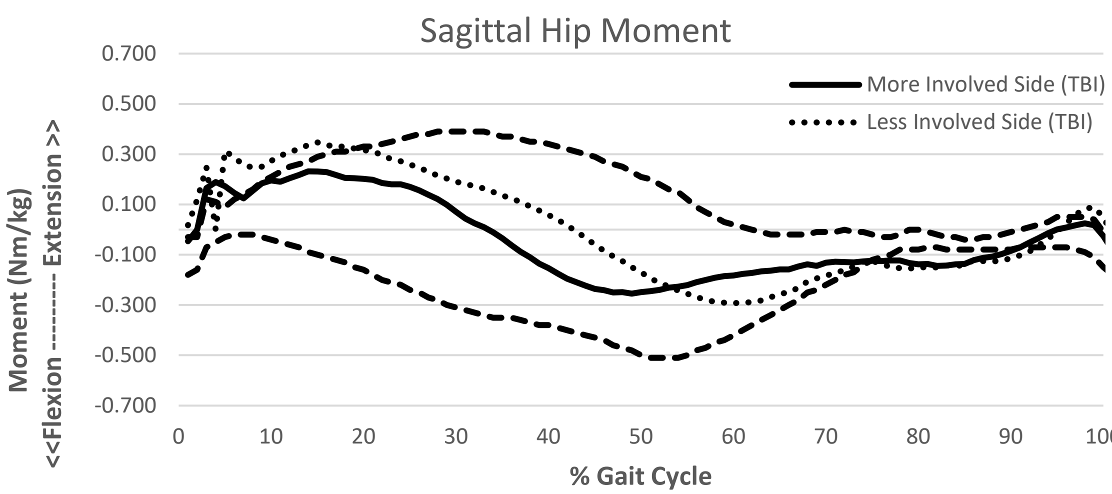

Figure 2. Sagittal Hip Moment of TBI Group Referenced to Velocity-Matched Normative Data

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ACKNOWLEDGMENTS

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DISCLOSURE STATEMENT

No authors have any conflicts of interest to disclose.

GAIT ANALYSIS OF SINGLE CASE WITH KNEE DISLOCATION AND POSTERIOR TIBIALIS TENDON TRANSFER SURGERIES WITH AND WITHOUT ORTHOSIS: BIOMECHANICAL CONSIDERATIONS

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PATIENT HISTORY

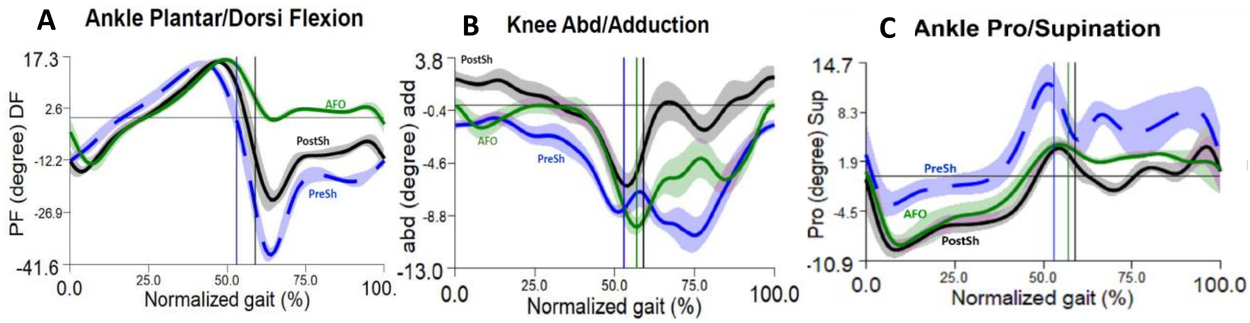
The subject is a 22-year-old female involved in a motor vehicle collision who sustained a lateral compression pelvic ring fracture, left knee dislocation, and severe L5 nerve root injury resulting in foot drop and weak hip abduction six months after the injury, multiligament reconstructive surgery was performed with allografts of both cruciate and both collateral ligaments (Schenck Classification KD4N injury reconstruction). Three months later (9 months post-injury), posterior tibial tendon transfer surgery (PTTTS)[1] and percutaneous lengthening of the Achilles tendon were performed to address drop foot gait deviation. A physical examination and gait analysis were conducted before and after the PTTTS. Gait was analyzed in three conditions: pre-surgery (Pre), post-surgery (Post) and post-surgery with an ankle-foot orthosis (PostAFO). This case report aims to emphasize the effectiveness of a surgeon-therapist collaboration to restore biomechanical function during gait by providing both surgical procedures and prescribed orthotics for a patient with a complex injury.

CLINICAL DATA

The subject was evaluated prior to- and 6 months after the PTTTS. Joint passive range of motion, single limb standing, and manual muscle testing (MMT) were conducted on her lower extremities by a licensed physical therapist. Significant findings include increases in strength from pre- to post-surgery of the left ankle dorsiflexors from 0/5 to 3-/5; knee flexors from 3+/5 to 4/5; hip abduction 3-/5 to 3+/5. Isometric strength of the left gastrocnemius, was measured using Biodex dynamometer decreased by 31.4%. The subject was unable to stand unsupported on her left single limb pre-surgery with only a minimal increase of 1.6 seconds post-surgery.

MOTION DATA

A 10-camera VICON motion capture system with three AMTI force plates were used to record the patient's gait pattern. The patient walked a 12m-pathway at self-selected speeds. The kinematic and kinetic data were calculated. The average of the 5 trials was used for analysis to compare the average data of each condition with published age-matched normative data. There was a significant improvement in swing time (Δ0.16s) from Pre to Post with increased ankle dorsiflexion (Δ14°) during the swing, while ankle flexion and extension were abnormal, perhaps due to gastrocnemius weakness (Figure A). There was a significant increase in gait speed from Pre to PostAFO (Δ0.18 m/s), which exceeded the clinical minimally important difference of 0.17 m/s.[2] Knee abduction motion increased with the PostAFO condition during terminal stance when compared to Post condition, which suggests increased stress on the knee with the addition of the AFO (Figure B). Additional findings include an increase in pronation during loading response from the Pre (3°) to the Post (10°) and PostAFO (9°) conditions. This is a significant increase when compared to normative data of 4° pronation[3,4] (Figure C).



Figures: A) Ankle Motion in the Sagittal Plane; indicating the improvement of drop-foot. B) Knee Motions in the Frontal Plane; indicating the improved loading patterns with AFO. The vertical lines in Figures A thru C are the end of stance phase. C) Pronation Motion during the stance phase.

TREATMENT DECISIONS AND INDICATIONS

The results confirmed the successful outcome of PTTTS to address the primary impairment of drop foot, increasing active ankle dorsiflexion with improved swing time on the left. Foot pronation during stance increased significantly post-surgically in both conditions. The use of the custom molded AFO did not improve the ankle pronation during the stance phase of gait. Additionally, use of the AFO resulted in an increase in knee abduction motion during terminal stance, when compared to no AFO post-surgery. One consideration would be to modify the AFO footplate to better support the foot-ankle complex and reduce the ankle pronation during stance. In addition, this adjustment may indirectly help reduce the abduction torque on the knee by better aligning the foot/ankle during heel off.[5] This finding indicates the clinical significance of proper AFO prescription, especially when designing a custom footplate, to better support the ankle-foot complex.[6] Due to the history of knee injury, excessive ankle pronation and continued gastrocnemius weakness, she is at risk of developing osteoarthritis.[7] As the subject continues to improve and her functional activity increases, frequent follow-up visits including re-evaluation of the AFO are needed to mitigate risk for osteoarthritis.

SUMMARY

This case study identifies the clinical significance of a properly fabricated orthosis for a patient with a multiligament knee injury and nerve dysfunction before and after PTTTS. Because some surgical techniques, like PTTTS, require the alteration of an anatomical structure and biomechanical function, an interdisciplinary approach, which include the physician, physical therapist and a biomechanical analysis can improve the plan of care and successful long-term outcomes.

DISCLOSURE STATEMENT All authors have no conflicts of interest to disclose.

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Evaluation of Gait and Spine Motion for a Pediatric Patient with Secondary Scoliosis: A Case Report

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PATIENT HISTORY

Patient is a 14 years old female who presented with primary ataxic cerebral palsy with secondary scoliosis. Patient also presented with signs of neurodegeneration with reduced depth perception, motor and language milestones delay, and partial agenesis of the corpus collosum on MRI studies.

CLINICAL DATA

Patient complained of difficulty ambulating. Her posture typically consisted of her head tilted to the right and rotating to the left. During gait she had a right lateral trunk lean in addition to the head positioning. The patient wished to ambulate with her spine and neck in neutral alignment.

MOTION DATA

The trunk lateral bend motion data during gait confirmed a right lateral flexion (Figure1&2). Furthermore, her pelvis was anteriorly tilted about 20 degrees and her hips were flexed greater than controlled baseline. Her pelvis was also retracted on the left and protracted on the right. Additional motion capture data was collected where the patient was instrumented with trunk markers to create a lumbar segment, thoracic segment and pelvic segment. She was instructed to stand still and bend maximally in all 3 directions (for/backwards, side-side, rotate L & R) to measure motion between these segments. During the standing trunk bending tests she seemed to present with reduced forward flexion (Table 1) motion compared to our previous published controls[1] suggesting restricted spine motion. She also showed an increased amount of out-of-plane motion (coronal & axial) during the forward bend suggesting structural or morphologic asymmetry. The asymmetry was further evidenced by the lateral bending tests (Table2).

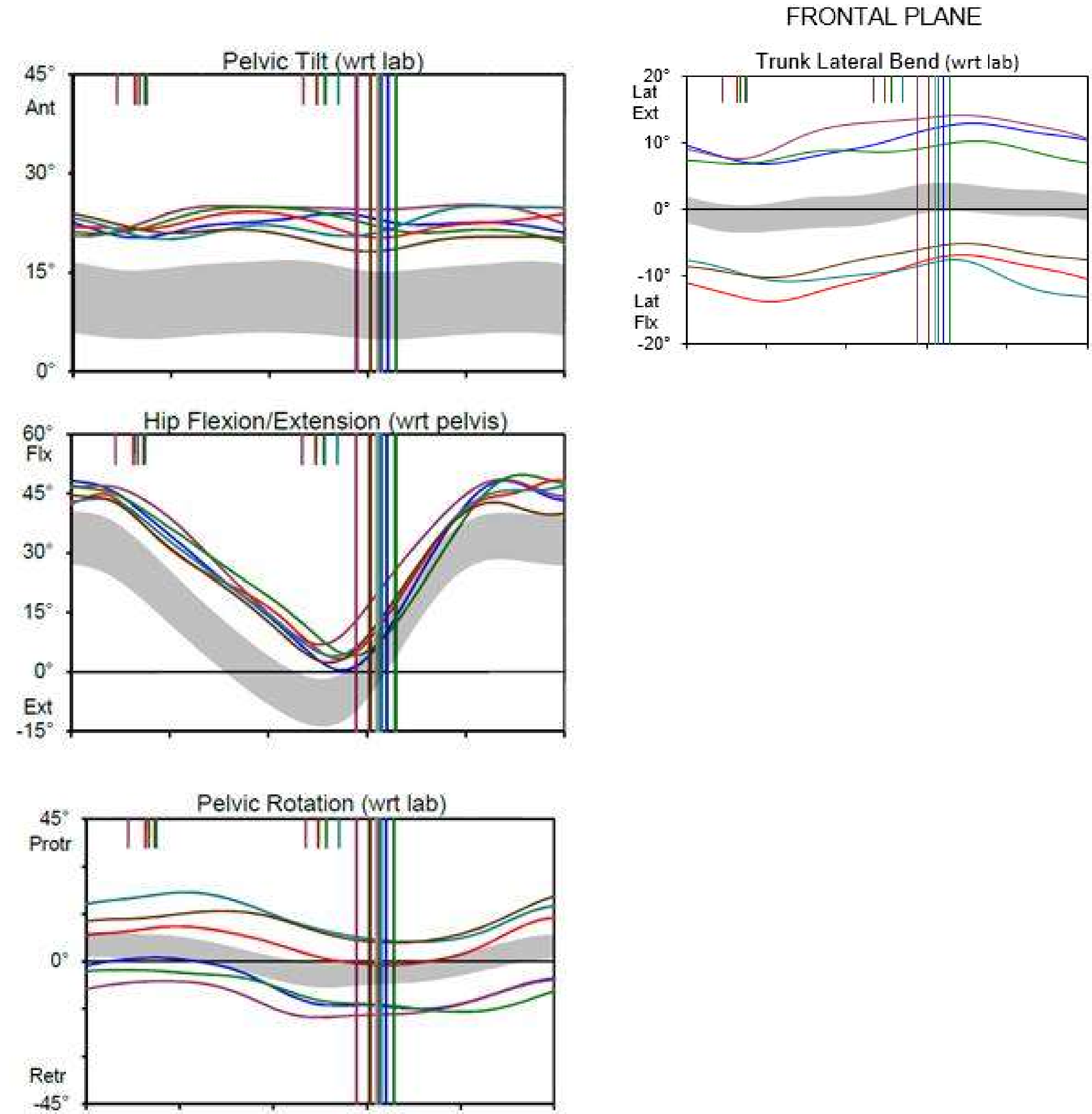

Figure 1. Gait analysis report on range of motion for pelvic tilt, hip rotation, hip flexion/extension, and trunk lateral bend

Figure 2. Patient's spine motion report for both lateral bends to left and right in comparison to baseline

TABLE 1: FLEXION STANDING ON FLOOR (ARMS BACK): (3 cycles)

FLX FLOOR	**Pre-Surgery**		
	Sagittal	Coronal	Axial
Thoracic (wrt lumbar)	**30° ±12.4**	9° ±1.1	4° ±1.9
Lumbar (wrt pelvis)	**45° ±2.3**	4° ±0.9	-1° ±1.0
Single Seg. (wrt pelvis)	**74° ±12.2**	18° ±1.8	11° ±1.1
Pelvis (wrt lab)	**20° ±5.4**	-5° ±1.4	-6° ±2.3
C7 to Floor %Δ in Vertical†:	**21%**		

TABLE 2: LATERAL BEND LEFT (ARMS CROSSED): (3 cycles)

LAT BEND L	**Pre-Surgery**		
	Sagittal	**Coronal**	Axial
Thoracic (wrt lumbar)	-16° ±2.3	**31° ±3.3**	11° ±0.3
Lumbar (wrt pelvis)	6° ±3.8	**12° ±0.8**	-3° ±4.1
Single Seg. (wrt pelvis)	-20° ±10.4	**38° ±1.1**	3° ±5.3
Pelvis (wrt lab)	13° ±1.4	**7° ±0.9**	38° ±1.7
L Shoulder to Floor %Δ in Vert†:	**10%**		

TREATMENT DECISIONS AND INDICATIONS

In this particular case, the patient's curve was about 60 degrees. Therefore, posterior spinal fusion is recommended due to the significant risk of progression. The goal of surgery would be to prevent scoliosis progression, rather than full correction. Surgery intervention corrected scoliosis from T4 to L4. Patient is currently wearing brace postoperatively and receiving post-operative rehabilitation.

OUTCOME

Patient is in 5 months post-operative recovery. We will plan to see her at 1-year post-operative to reassess improvements. Patient had a post T3-L4 posterior spinal and is currently in rehabilitation training for her posture.

SUMMARY

The current case study describes the successful surgical management of a patient with scoliosis secondary to ataxic cerebral palsy. Post-surgery recovery is currently undergoing and has shown promising recovery and improvements in waist range of motion, gait stability, function, and participation.

DISCLOSURE STATEMENT

We have no conflicts of interest to disclose.

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EXPLORING DYNAMIC ELECTROMYOGRAPHY AS A BIOMARKER: A CASE STUDY OF AN ADOLESCENT WITH CMT TYPE 2

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PATIENT HISTORY

A 17-year-old female, with a diagnosis of CMT Type 2, was the product of a full term birth and began walking at 1 year of age. She was initially referred for a motion analysis at 7 years of age (Test 1) for orthopedic surgical decision-making to address bilateral foot and ankle instability and positioning, bilaterally. Parents reported functional issues related to CMT including reduced walking velocity and tripping and falling. Although she demonstrated improvements with her bilateral ankle-foot orthoses, surgical intervention was proposed to maintain foot position for comfortable bracing. She returned 4 years following (Test 2) for a post-operative gait analysis and then again 3 years later (Test 3) to evaluate for ongoing gait changes as part of a research protocol.

CLINICAL DATA

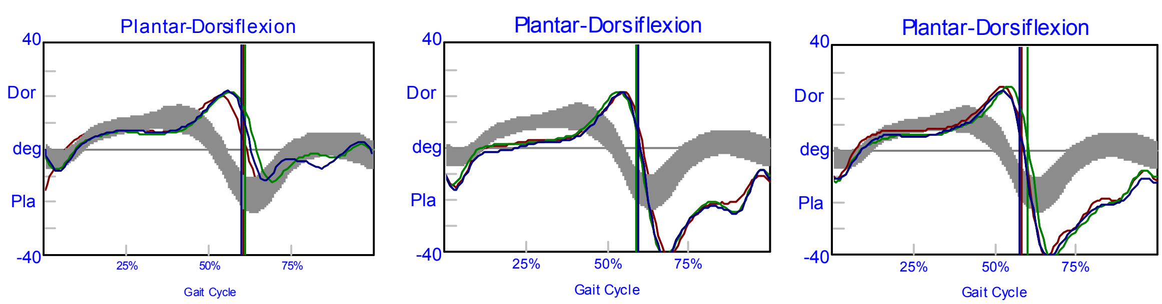
A comprehensive gait analysis was completed including 3D gait analysis barefoot and AFO walking and dynamic electromyography (dEMG) during barefoot walking at all three assessments. Selected clinical exam findings are summarized in Table 1 and show decline in strength over time.

Table 1: Selected clinical exam and temporal findings for pre and post-surgery.

Gait analysis test	**Test 1**	**Test 2**	**Test 3**
Ankle dorsiflexion strength (right side)	4/5	1/5	0/5
Ankle plantar flexion strength	2/5	2/5	0/5
Walking velocity – barefoot (m/sec)	1.21 – 1.33	1.23 – 1.32	1.10 – 1.19

MOTION AND dEMG DATA

At the initial gait analysis, ankle sagittal plane motion showed findings consistent with CMT with increased and delayed peak dorsiflexion in stance with ability to dorsiflex in swing (Fig. 1a). Dynamic EMG data showed activity in mid swing for the right anterior tibialis.



a) Pre-op b) Post-op 1 c) Post-op 2

Figure 1: Right side (3 gait cycles) sagittal plane ankle kinematics during walking.

TREATMENT DECISION AND INDICATIONS

Bilateral orthopedic surgery was staged at 1.5 and 2.5 years following the gait analysis: 1) calcaneal osteotomy to correct the hindfoot varus and cavus, 2) anterior tibialis transfer to the mid foot to correct forefoot supination in swing and at initial contact, decreased internal foot progression in stance, and delay risk of recurrence as patient had normal ankle dorsiflexion in swing, 4/5 strength and appropriate anterior tibialis EMG in swing and 3) posterior tibialis lengthening to reduce increased forefoot adductus and internal foot progression.

Table 2: Comparison of dEMG data changes over time for the right anterior tibialis.

	TD	**CMT Type I**		
Gait analysis test	n/a	**Pre-op**	**Post-op 1**	**Post-op 2**
iEMG (v)	812.4	454.5	134.5	28.2
RMS (v)	0.89	0.38	0.12	0.03
Mean Frequency (Hz)	137.6	149.7	120.5	90.8
Median Frequency (Hz)	114.2	128.9	94.1	63.1
Mean Power Frequency (Hz)	281.1	194.8	136.7	29.2
Median Power Frequency (Hz)	262.4	165.6	116.1	3.3

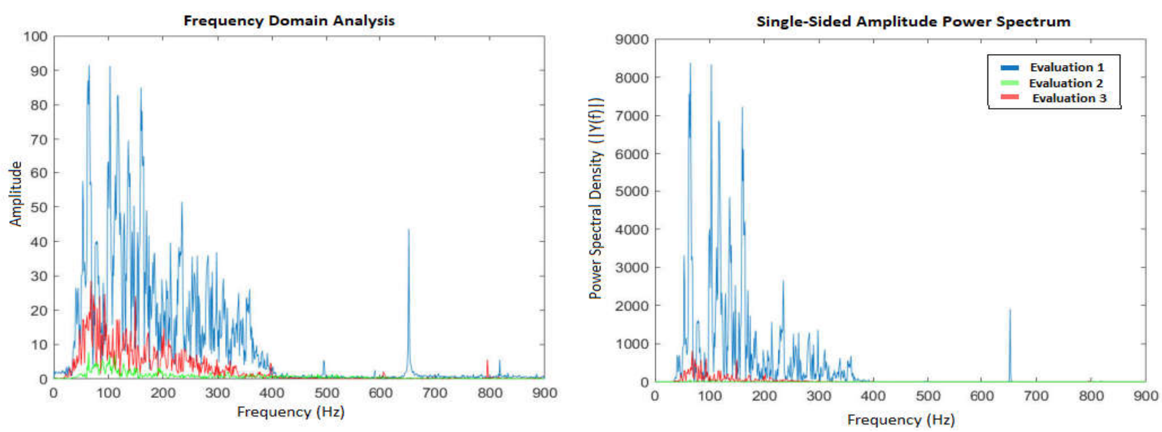


Figure 2: Frequency and power analyses for the right anterior tibialis shows substantial change from Test 1 to 2.

TREATMENT OUTCOMES

At post-op 1, the right ankle sagittal plane kinematic showed similar peak dorsiflexion in terminal stance, however, there was excessive equinus in swing (Fig. 1b). These findings were similar at the post-op 2 test (Fig. 1c). A retrospective analysis of the dEMG signals using additional measures including iEMG, RMS and frequency domain for the anterior tibialis showed values that differed from typically developing (TD) for the anterior tibialis at pre-op, that declined from pre-op to post-op 1 and then declined again at post-op 2. (Table 2).

SUMMARY

Prior to surgery, the anterior tibialis function showed the onset of weakness on clinical assessment, however, during barefoot gait the patient was still able to dorsiflex the ankle for clearance and therefore, surgical intervention was deemed appropriate. The dEMG findings however, showed differences from TD at pre-op that were not yet manifest in the ankle kinematics. It is possible that the initial findings from the dEMG measures are predictive of pending decline in muscle function and should guide future treatment decisions in terms of understanding possible prognosis for treatment outcomes. Dynamic EMG may also have the potential to be a biomarker so that pending therapeutic treatments can be started as early as possible.

DISCLOSURE STATEMENT

The authors have no conflicts of interest to disclose.

The Five Year Outcome of the Ponseti Method in Children with Idiopathic Clubfoot and Arthrogryposis

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Introduction
Clubfoot presents as a foot with cavus, forefoot adduction, and equinovarus [1]. Arthrogryposis is a condition that manifests in multiple joint contractures and muscle weakness [2]. Most children with arthrogryposis have bilateral clubfeet, and their clubfeet are often much stiffer and more rigid than idiopathic clubfeet [2]. In this study, we compared the effectiveness of the Ponseti method over a five year span in treating children with idiopathic clubfoot and clubfoot associated with arthrogryposis.

Clinical Significance
Idiopathic clubfoot has shown to be corrected with a nonoperative approach called the Ponseti method [1]. Previous treatment of the stiffer clubfoot associated with arthrogryposis relied heavily on surgical intervention, which led to scar tissue, pain, and a high rate of relapse [2]. The recent clinical trend is to move to the more conservative Ponseti method to treat the clubfeet in children with arthrogryposis, but the research to support this is limited [2].

Methods
This retrospective review was determined exempt by our institutional IRB policy. In order to be included, children must have visited the gait lab between the years 2012-2019 with a diagnosis of idiopathic clubfoot or clubfoot associated with arthrogryposis. Children must have been between the ages of 4.9-6.9 years of age when evaluated, and must have been treated with the Ponseti method. Children with nonidiopathic clubfeet or congenital vertical talus were excluded. Data from the gait lab included passive range of motion measurements, foot pressure percentages, Gross Motor Functional Measure Dimension D (GMFM-D) scores, and Pediatric Outcomes Data Collection Instrument (PODCI) scores. Children with idiopathic clubfoot and arthrogryposis were compared to typically developing peers. Typically developing data was retrieved from prior gait lab studies or published journal standards.

Results
This study included a total of 117 children, 28 children with arthrogryposis (56 feet; all bilaterally involved) and 89 children with clubfoot (134 feet; 45 children bilaterally involved, 44 children unilaterally involved). The average age of these two groups was 4.8 ± 0.8 years. Our results showed a residual equinovarus deformity in both the arthrogryposis and clubfoot group when compared to their typically developing peers (Table 1). The equinus deformity led to reduced heel pressure and early time to heel rise values. The residual varus deformity was seen in high lateral midfoot pressure (LMF), low medial forefoot pressure (MFF), and negative CPPI. Children with idiopathic clubfoot and clubfoot associated with arthrogryposis showed residual deformity in their range of motion, specifically their limitations in dorsiflexion and plantarflexion. When the idiopathic clubfoot group was compared to the arthrogryposis group there was significantly more limited gross motor function in the arthrogryposis group according to their GMFM-D scores [3]. Children with arthrogryposis demonstrated significantly more

limited functional ability (PODCI parent report and GMFM-D) and the presence of pain, while the idiopathic clubfoot group showed a higher functional ability and no pain. The children with idiopathic clubfoot and the children with arthrogryposis were statistically different in all six domains of the PODCI. When assessed functionally, the arthrogryposis children showed greater limitations when compared to the idiopathic clubfoot and typically developing group.

Table 1. Foot pressure percentages, passive range of motion, gross motor functional measure scores, and parent reported outcome survey scores for children with idiopathic clubfoot (clubfoot/C), children with clubfoot associated with arthrogryposis (arthrogryposis/A), and typically developing children (typically developing/TD). Significant results as per Hochberg correction to Welch's t-test are shown with an asterisk.

	Arthrogryposis (A)	Clubfoot (C)	Typically Developing (TD)	A vs. C	A vs. TD	C vs. TD
Foot Pressure						
Heel %	29 ± 26	29 ± 17	41 ± 10	0.94	0.089	0.000018*
Time to Heel Rise	48 ± 29	48 ± 17	57 ± 9	0.99	0.23	0.00056*
LMF %	38 ± 28	29 ± 15	10 ± 7	0.22	0.0009*	3.80e-15*
MFF %	4 ± 8	3 ± 3	1 ± 1	0.47	0.130	0.0004*
CPPI	-41 ± 51	-38 ± 34	-5 ± 35	0.85	0.019*	0.00021*
PROM						
Dorsiflexion	-5 ± 13	1 ± 8	11 ± 8	0.11	0.00022*	0.000003*
Plantarflexion	33 ± 18	42 ±11	65 ± 5	0.048	0.0000024*	1.10e-22*
GMFM-D	27 ± 11	36 ± 2	n/a	0.000087*	n/a	n/a
PODCI						
Transfer/Basic Mobility	78 ± 19	96 ± 6	99 ± 3	0.000059*	0.0000092*	0.013
Pain	82 ± 19	91 ± 13	95 ± 18	0.025*	0.035	0.47
Global Function	72 ± 16	92 ± 8	95 ± 7	0.00000077*	0.0000001*	0.11

Discussion

The Ponseti method is effective in treating clubfoot, both idiopathic and related to arthrogryposis [2]. Some residual equinovarus was found in the idiopathic clubfoot group, but these children were found to be highly functional with no significant complaints of pain. The children with arthrogryposis also had residual deformity, as well as significant functional limitations, but with conservative management a braceable foot was created without invasive treatment intervention.

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Disclosure Statement: No conflicts of interest to disclose

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Correlation between Foot Posture Index and Radiographic Parameters

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INTRODUCTION

The Foot Posture Index (FPI-6) is a validated and rapid clinical method for evaluating standing foot posture through assessing six individual criteria. However, evidences for correlation of FPI with radiographic measurements are still scanty. So the objective of this study is to investigate the correlation of FPI with radiographic measurements and to assess feasibility of FPI-6 for clinical evaluation of standing foot postures.

CLINICAL SIGNIFICANCE

By elucidating the correlation of FPI-6 with radiographic measurements, we might provide the clue for whether FPI-6 will be an appropriate tool for clinical application of foot postures in the absence of radiographic examinations

METHODS

Sixty patients (M:F,33:27, mean age 62) who visited our hospital for treatment of foot and ankle symptoms and 40 asymptomatic male volunteers (age, 20-28) were included in this study. FPI scores were evaluated by 4 raters and inter-rater reliability of FPI was evaluated by intraclass correlation coefficient (ICC). Radiologic measurements including Meary's angle (MA), talonavicular coverage angle (TNCA), talocalcaneal angle (TCA), calcaneal pitch angle (CPA), and hindfoot alignment angle (HAA) were measured. For correlation analysis between FPI-6 and radiographic measurements, average FPI-6 values of two most experienced examiners was selected and evaluated using Pearson's correlation analysis.

RESULTS

In general, inter-rater reliability was 'high to excellent' among raters. However, FPI-6 score was more repeatable in pathologic conditions and in more experienced raters. ICC of FPI-6 of all participants was 0.927 in two most experienced examiners. FPI-6 score ranged from -12 to +12 in patients group, and ranged from -3 to +10 (average, 0.937) in asymptomatic group.

In patients and symptomless volunteers, FPI-6 score was correlated with TNCA, AP talo-1st metatarsal angle, lateral TCA, MA, and HAA. Especially, TNCA and HAA was more strongly correlated with FPI-6 scores. When we analyzed subdomain of FPI scoring system, congruence of the medial longitudinal arch was the most powerful indicator for TNCA and prominence in the region of the talonavicular joint was for HAA.

Table. Correlation analysis results of total 100 participants

	TCA	TNCA	Talo1st MTA	TCA lat	CPA	Meary A	HAA
FPI 1	0.136	0.620*	0.492*	0.399*	-0.052	0.530*	-0.713*
FPI 2	0.109	0.602*	0.398*	0.317*	-0.066	0.505*	-0.738*
FPI 3	0.127	0.583**	0.339**	0.356**	0.002	0.467*	-0.726**
FPI 4	0.072	0.573**	0.358**	0.323**	-0.052	0.485**	-0.748**
FPI 5	0.166	0.686**	0.438**	0.318**	-.263**	0.658**	-0.664**
FPI 6	0.168	0.597**	0.503**	0.346**	-0.095	0.512**	-0.658**
Total FPI	0.140	0.665**	0.453**	0.369**	-0.096	0.570**	-0.773**

Note: Data results are presented as correlation coefficient between the FPI-6 and radiographic variables; Pearson correlation matrix
*$p<0.05$ **$p<0.001$

DISCUSSION

Although experience of raters does affect FPI-6 scores, repeatability of FPI assessment was substantial and there was some correlations between FPI and radiographic measurements in foot and ankle patients. FPI scores were correlated with several radiographic parameters in weak correlation in asymptomatic volunteers, but more strongly in patients with radiographic deformity. We think FPI-6 scoring system can be used as a first line of foot posture assessment in clinics without radiographic evaluation if used carefully.

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DISCLOSURE STATEMENT

The authors have no conflicts of interest to disclose.

The difference of in-shoe plantar pressure between level walking and stair walking

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INTRODUCTION

Stair walking is one of common activities of daily living. It is more demanding than level walking and can aggravate discomfort of the foot, such as Morton's neuroma, plantar fasciitis, Achilles tendinitis, pressure related-ulcer, and etc. Therefore, analysis of increased pressure in specific plantar area at stair walking can be used as a risk assessment of foot discomfort and basic data in the clinical field. The purpose of this study is to analyze plantar pressure distribution and pressure patterns during gait cycle at stair walking compared to level walking.

CLINICAL SIGNIFICANCE

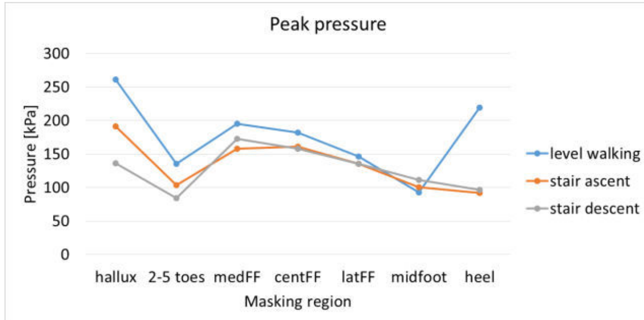
By analyzing plantar pressure pattern and distribution in stair walking, we might elicit that change of daily living pattern would be important for preventions of foot disorders related with plantar pressure.

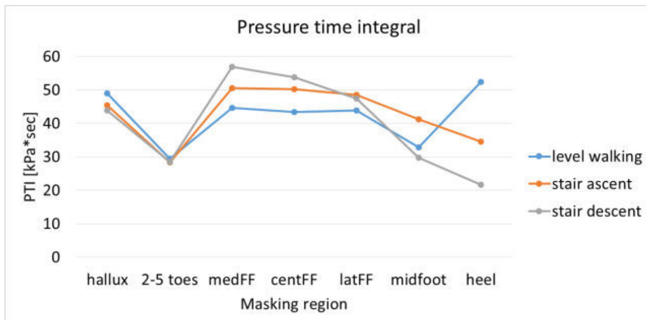
METHODS

Thirty five healthy males with 20-28 years old were included in this study after examining normality. They performed level walking, stair ascending, and descending in same type of shoes. Measurements of in-shoe plantar pressure including peak pressure (PP), pressure-time integral (PTI) were done by Pedar-X system, masked 7 segments in sole. Percentages were assigned in relation to the size for each mask segment. Lastly, pressures in each region throughout the gait cycle were extracted from each type of walking. Statistical analysis was performed using repeated measure ANOVA, and Bonferroni post hoc test was done.

RESULTS

PP in all regions except for the midfoot were higher during level walking than stair walking. During stair descent, mean peak pressures in all the regions except for the midfoot were generally lower than other types of walking, but it was the highest in the midfoot region. PTI in the medial and central forefoot was higher during stair descent than level walking. PTI in the central and lateral forefoot, and the midfoot was higher when stair ascending than level walking. PTI in the heel region was the highest during level walking, followed by stair ascent, stair descent.





DISCUSSION

The risk of aggravation of discomfort in the midfoot area increases when stair descending. The medial region of forefoot bear high pressure load during stair descent, and the lateral region of forefoot and the midfoot region bear high pressure load during stair ascent. This is the first study to show plantar pressure patterns during level and stair walking in the large healthy gender-controlled population. We recommend that patients with pressure related foot lesions in the forefoot or midfoot avoid stair walking.

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ACKNOWLEDGEMENTS

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DISCLOSURE STATEMENT

The authors have no conflicts of interest to disclose.

INTERSEGMENTAL FOOT MOTION BEFORE AND AFTER HIGH TIBIAL OSTEOTOMY IN GENU VARUM PATIENTS

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INTRODUCTION

High tibial osteotomy (HTO) is a well-established treatment for medial compartment knee osteoarthritis with genu varum [1]. This treatment shifts the weight-bearing axis from the medial to the lateral side of the knee [2]. As a result, the load of the body is directly concentrated on the ankle joint. This has caused some concerns about the effect of knee osteoarthritis on the foot and ankle. However, to the best of our knowledge, no studies using foot gait analysis have been performed to evaluate the intersegmental motions of these patients.

CLINICAL SIGNIFICANCE

This study highlights the difference in intersegmental foot motion before and after high tibial osteotomy in genu varum patients.

METHODS

The study included twenty-four patients with medial compartment knee osteoarthritis with genu varum who underwent HTO between March 2012 and April 2016. The follow-up period for patients was at least twelve months after HTO. The control group consisted of forty-eight elderly participants who were recruited from local volunteers. Segmental foot kinematics were evaluated using a 3D multi-segment foot model (DuPont foot model) and gait data were collected using twelve cameras. The data of temporal and spatial gait parameters were obtained. Statistical analysis was performed.

RESULTS

There was a tendency for gait speed to increase after HTO. Normalized stride length was significantly increased after HTO. In hallux kinematics relative to the forefoot, sagittal motions of patients and the control group were similar to each other throughout most of the gait cycle (Fig. 1). In forefoot kinematics relative to the hindfoot, the pre-HTO state showed significant pronation throughout the gait cycle, while the post-HTO state showed similar position and motion with normal control. In hindfoot kinematics relative to the tibia, coronal motions of the pre-HTO state showed supination throughout the gait cycle, while supination during the stance phase decreased after HTO.

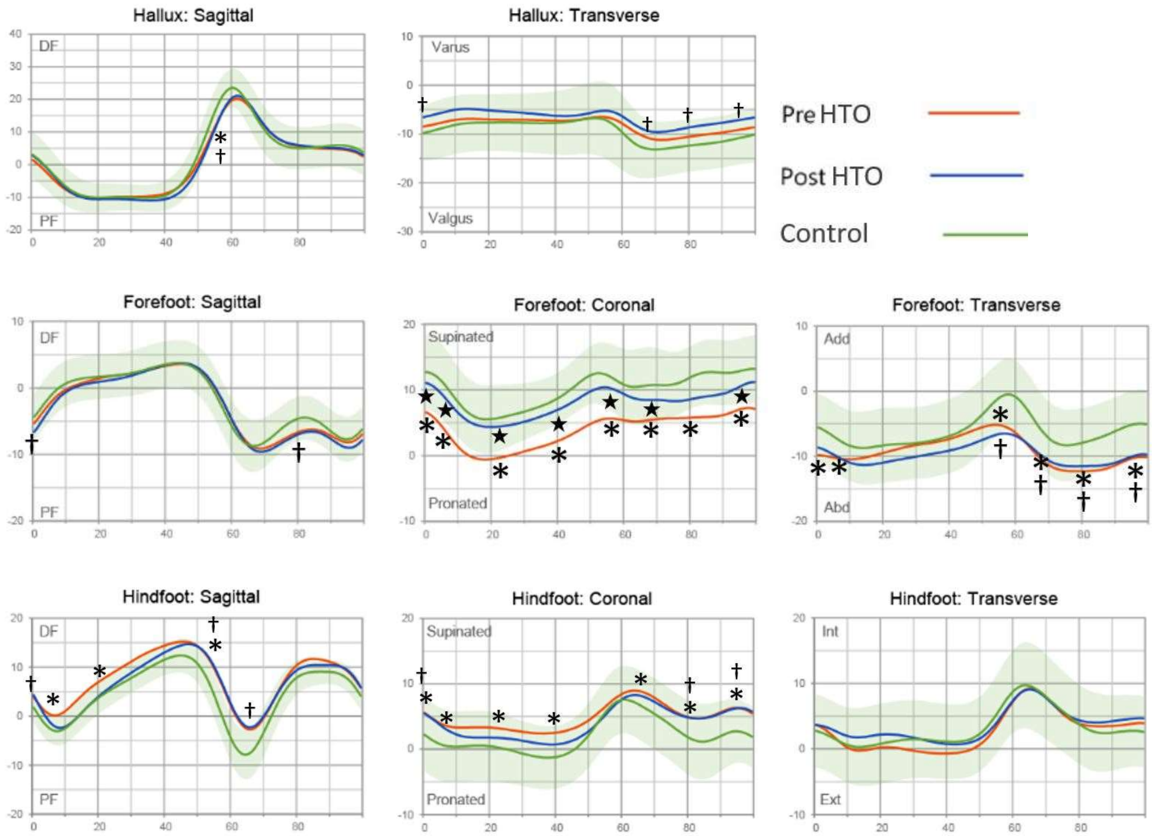

Figure 1: Average kinematics of the foot during the whole gait cycle

DISCUSSION

Genu varum patients with medial compartment knee osteoarthritis showed different gait parameters and intersegmental motion during gait when compared with elderly, gender matched controls. Furthermore, the effect of HTO was demonstrated by the loss of midfoot compensation, where the midfoot was pronated to attain plantigrade gait in genu varum patients. As segmental motions of the foot and ankle are affected by the adjacent knee joint, attention should be given to the potential for osteoarthritis in the foot and ankle when treating patients with knee osteoarthritis.

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DISCLOSURE STATEMENT

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COMPARING THE KINEMATICS, REPEATABILITIES AND REPRODUCIBILITIES OF FIVE MULTI-SEGMENT FOOT MODELS BASED ON DIFFERENT ANALYSIS METHODS

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INTRODUCTION

Multi-segment foot models for assessing three-dimensional foot motion are analyzed via either the marker-based method or the bone-based method. The aim of this study is to compare the kinematics, repeatabilities, and reproducibilities of five popular multi-segment foot models when walking with the markers of the models.

CLINICAL SIGNIFICANCE

This study offers researchers or clinicians the information of differences in multi-segment foot models that generally used in clinical laboratory.

METHODS

Eleven healthy males with a mean age of 26.5 years participated in this study. We created a merged 29-marker set including three marker-based models: Oxford (OFM) [1], modified Rizzoli (mRFM) [2], and DuPont (DFM) [3] and two bone- and marker-based models: Milwaukee (MFM) [4] and modified Shriners Hospital for Children Greenville (mSHCG) [5]. Two operators applied the merged model to participants two times and divided the segments into shank, hindfoot, and forefoot. Coefficients of multiple correlation (CMC) were used to assess both repeatability and reproducibility, and statistical parametric mapping (SPM) of the t-value was employed for kinematics comparison.

RESULTS

In the sagittal plane, the CMC values of all segments were greater than 0.840 in all models. However, for the hindfoot motion in the coronal plane, the CMCs of MFM and mSHCG were greater than 0.793, whereas those of the other models were less than 0.660. MFM and mSHCG showed more dorsiflexed hindfoot and plantarflexed forefoot compared to the other models. There were no significant differences in hindfoot varus/valgus among the models and in all hindfoot kinematics of OFM and DFM. DFM and mSHCG tended to have a higher ROM than the other models.

Table 1: CMC values of repeatability (intra- and inter-sessions) and reproducibility (inter-rater) of relative motions of the shank-hindfoot and hindfoot-forefoot.

Model	Plane	Shank - Hindfoot				Hindfoot - Forefoot			
		Intra-session	Inter-session		Inter-rater	Intra-session	Inter-session		Inter-rater
			rater A	rater B			rater A	rater B	
DFM	Sagittal	0.959	0.903	0.950	0.934	0.967	0.869	0.958	0.920
	Coronal	0.926	0.554	0.584	0.652	0.929	0.696	0.606	0.733
	Transverse	0.895	0.461	0.777	0.627	0.886	0.618	0.850	0.837
mRFM	Sagittal	0.957	0.957	0.952	0.944	0.983	0.971	0.947	0.929
	Coronal	0.923	0.194	0.660	0.401	0.931	0.862	0.835	0.862
	Transverse	0.943	0.807	0.838	0.778	0.968	0.746	0.723	0.332
OFM	Sagittal	0.966	0.946	0.960	0.962	0.982	0.934	0.941	0.924
	Coronal	0.923	0.327	0.520	0.410	0.961	0.481	0.453	0.612
	Transverse	0.953	0.752	0.867	0.791	0.988	0.931	0.930	0.846
MFM	Sagittal	0.968	0.933	0.946	0.952	0.987	0.968	0.978	0.965
	Coronal	0.940	0.859	0.793	0.898	0.902	0.702	0.771	0.848
	Transverse	0.971	0.939	0.654	0.728	0.936	0.873	0.931	0.628
mSHCG	Sagittal	0.962	0.960	0.962	0.971	0.981	0.926	0.840	0.871
	Coronal	0.929	0.836	0.897	0.742	0.924	0.873	0.663	0.720
	Transverse	0.946	0.734	0.881	0.827	0.974	0.672	0.797	0.441

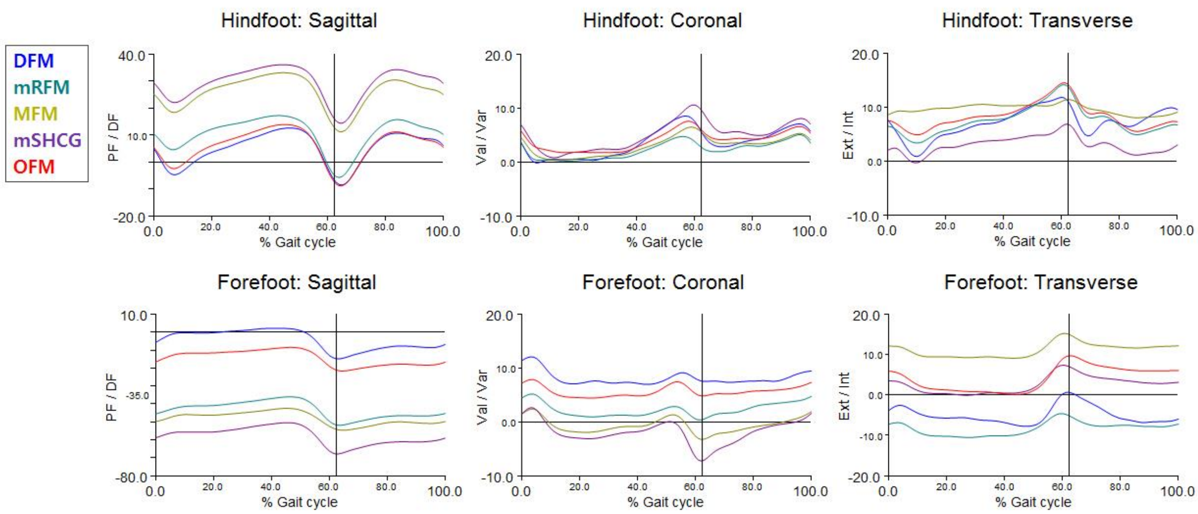

Figure 1. Average walking kinematics for five models (% gait cycle)

DISCUSSION

High repeatability and reproducibility in the sagittal plane were verified. In general, the bone- and marker-based models were more reliable, especially in the hindfoot kinematics. All models had different initial angles, and it were noticeable in the sagittal plane. Consequently, it was important when comparing the clinical papers used in different foot models on patients with the same diseases.

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DISCLOSURE STATEMENT

The authors have no conflicts of interest to disclose.

THE EFFECT OF MILD TO MODERATE JUVENILE HAllUX VALGUS ON GAIT

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INTRODUCTION

Juvenile hallux valgus (HV) is a common deformity in clinics. The HV deformity is defined as lateral deviation of the first toe and medial deviation of the first metatarsal with the apex of the deformity at the first metatarsophalangeal joint.

Previous studies on patients with HV have focused mainly on the radiographic assessment and the plantar pressure distributions. Actually, it affects not only the foot itself, but also the other lower limb joints. Kao-Shang Shih[1] in Taiwan did the research about adult patients with HV by three-dimensional gait analysis. The research showed that the HV group had some compensatory changes, primarily reduced toe-out angle, increased hip internal rotation angle and knee abductor moments. This strategy was to reduce the pain on the first metatarsophalangeal joint when patients were walking.

Children with HV usually don't feel painful. So the gait changes in juvenile hallux valgus may be different with adult patients. Different degrees of hallux valgus angle may have different effects on gait.

This study focused on the effect of mild to moderate Juvenile hallux valgus on gait.

CLINICAL SIGNIFICANCE

The purpose of the study was to investigate the kinematic and kinetic changes of the lower limb joints during level walking in children with HV. Children with mild to moderate hallux valgus are not getting much attention. This study can provide clinicians with information about the gait characteristics of mild to moderate juvenile hallux valgus and can help clinicians develop nonsurgical treatment plans for these children.

METHODS

20 children with bilateral HV （age: 8.8±1.5 years, mass: 33.2±8.9kg, height: 139.4±10.4cm, HVA: 22.7±4.4°）and 20 healthy children controls (age: 8.8±1.4 years，mass: 33.2±9.2kg, height: 138.2±9.4cm) walked while three-dimensional kinematic and kinetic data were measured (three-dimensional gait analysis system, 19 cameras from Motion, 4 force plates from Bertec, USA). Kinematic and kinetic data included moment, power, ground reaction force, angle of hip, knee and ankle joint, temporal-distance parameters and so on. Comparisons of the variables between the HV and control groups were performed using independent t-tests.

RESULTS

No significant differences were found between the two groups in age, body height and mass. No statistical difference were found in toe out angle and hip internal rotation angle.

Patients with HV showed significantly reduced forward velocity. The HV group also showed significantly reduced ground reaction force and hip extension angle at toe off.

DISCUSSION

There was no statistical difference in the toe out angle and hip internal rotation angle indicated that children with mild to moderate hallux valgus did not have the pain-avoidance mechanism in adult patients during walking. So their gait patterns were different. However, the decreased hip extension angle and ground reaction force at toe off indicated that the abnormal alignment of the great toe reduced the force to push off the ground, leading to the inadequate hip extension angle, thus reducing the walking speed. In conclusion, the effect on gait in children with mild to moderate hallux valgus is mainly manifested in the decreased ability of feet to push off the ground and the effect of walking efficiency.

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DISCLOSURE STATEMENT

The authors declare that they have no conflicts of interest to disclose.

DEVELOPMENT OF A KINETIC SEGMENTAL FOOT MODEL

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INTRODUCTION

Determination of hindfoot, forefoot, and hallux kinetics is important in clinical gait analysis. Several current multi-segmental foot models have the capability to calculate segmental kinetics, however, the complexity of the data collection for these models limits their utility in clinical settings [1-2]. The purpose of this work is to develop a mathematical algorithm capable of distributing the single composite ground reaction force (GRF) amongst multiple foot segments. Accurate distribution will allow for segmental kinetics to be determined without the need for additional equipment, additional time in the gait lab, or targeted walking by patients.

CLINICAL SIGNIFICANCE

Understanding the biomechanics of the foot is critical to the proper care of patients with a variety of orthopaedic impairments (e.g., cerebral palsy, Charcot-Marie-Tooth disease) and foot deformities, such as clubfoot, equinovarus, and pes planovalgus. Accurate knowledge of three-dimensional (3D) hindfoot, forefoot, and hallux kinetics during functional activities such as walking is essential for identifying the effects of such injury and disease.

METHODS

An inverse dynamics approach, based on Vaughan's lower extremity gait analysis, was used to determine the 3D kinetics at the metatarsophalangeal (MTP) (hallux relative to forefoot), mid-tarsal (forefoot relative to hindfoot), and ankle (hindfoot relative to tibia) joints [3]. Forces and moments were calculated at each joint starting distally with the hallux segment and progressing proximally using Newton-Euler equations. Motion capture data was collected using the previously validated kinematic Milwaukee Foot Model (MFM) [4] for the shank, hindfoot, forefoot, and hallux segments. A series of radiographs was taken in accordance with the MFM procedure. Inertial segmental properties were calculated using CT scans of a control foot. Local inertial matrices were determined in the control foot by estimating the segments as elliptical cylinders. Subject specific properties were scaled to the control foot based upon the subject's height, mass, and foot marker positions in the static pose.

The distribution of the GRF is based upon the three phases in existing models used in induced acceleration analyses. In these models, kinematic constraints are placed on the foot as a function of the center of pressure (COP) as developed by Lin et al (Figure 1) [5]. Phase one, corresponding to heel strike, occurs when the location of the COP is posterior to the axis created by the medial and lateral calcaneus (Axis 1). At this phase, the GRF is acting only on the hindfoot segment. Phase two, corresponding to foot flat, begins as the COP moves past Axis 1. At this point the GRF begins to distribute between the hindfoot and forefoot segments based on the following equations:

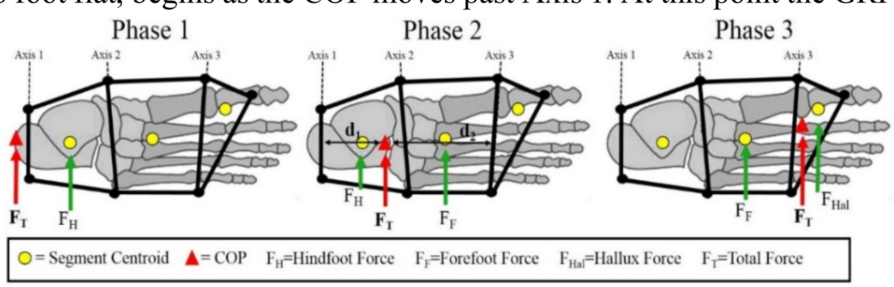

Figure 1: GRF distribution based on COP

$$w = \frac{d1}{d1 + d2}, \qquad F_H(t) = (1 - w) * F_T, \qquad F_F(t) = w * F_T,$$

where d_1 is the shortest distance from Axis 1 to the COP and d_2 is the shortest distance from Axis 3 to the COP. As the COP reaches Axis 3, and d_2 approaches zero, the entire GRF is located on the forefoot. Phase three, corresponding to toe off, occurs as the COP moves past Axis 3. In this phase there is no component of the GRF on the hindfoot segment and the GRF is distributed between the forefoot and hallux in the same manner as described for phase two.

Evaluation of the GRF modeling technique was also completed using an RSscan pedobarographic pressure platform [2]. The pressure platform was synced with the AMTI force plate to allow for simultaneous collection of force plate and pressure plate data.

DEMONSTRATION

Motion capture data, for a 16 year-old male with no reported foot pathology, was collected simultaneously using both an AMTI force plate and an RSscan pedobarographic plate. The force distribution throughout stance was calculated using the mathematical algorithm and the pedobarographic distribution. The moments at the joints were calculated using both force distributions (Figure 2).

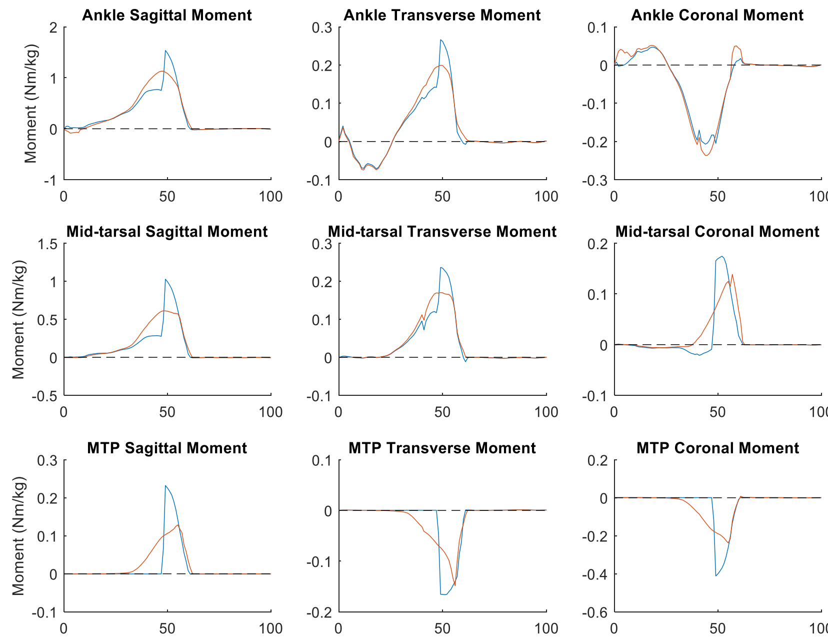

Figure 2: Normalized Joint Moments over 100% gait cycle. *Blue: mathematical algorithm; Red: RSscan pedobarographic data*

SUMMARY

This work demonstrates the feasibility of obtaining segmental foot kinetics using a single force plate which is standard equipment in clinical gait labs. Our algorithm results show good agreement with the pedobarographic platform, especially with the hindfoot force distribution, resulting in closely matched moment plots in all three planes of ankle motion. Modifications to the force division algorithm between the forefoot and hallux segments are in process, in order to improve moment results at the mid-tarsal and MTP joints. Future work is being directed towards further evaluation of the system and algorithms with a larger population in order to better quantify typically developing kinetics at the ankle, mid-tarsal, and MTP joints. Identifying kinetic corridors for the moments at the ankle, mid-tarsal, and MTP joints will allow for better comparison and discovery of differences in pathological feet, leading to better management of gait challenges.

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DISCLOSURE STATEMENT - We have no conflicts of interest to disclose.

EFFECT OF KINESIO TAPING COMBINED WITH MUSCLE TRAINING ON CHILDREN WITH FLEXIBLE FLATFOOT

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INTRODUCTION

Influenced by age, weight, ligament relaxation and imbalance of muscle strength, flexible flatfoot are most common in children with developmental abnormal alignment of force lines[1]. The collapse of medial longitudinal arch of foot increases abnormal stress of foot, bones and joints, and affects children's motor function. Using insole is the main treatment way of flexible flatfoot. But some studies suggest that the effect of insole is not accurate[2].The insole also has the problem that it cannot work on children without shoes. Kinesio taping method has been proved to have the functions of increasing proprioception and enhancing muscle activity[3]. Compared with insoles, it is more comfortable and convenient.

This study aimed to investigate whether that use of Kinesio taping combined with foot muscle training could improve the gait function and foot posture of children with flexible flatfoot.

CLINICAL SIGNIFICANCE

To investigate the effectiveness of Kinesio Taping combined with muscle training on kids with flexible flatfoot.

METHODS

Sixteen males (29 feet) and thirteen females (22 feet) at an average age of 5 years (S.D.1.84; range 4-11) and BMI15.94 kg/m (S.D.1.65) with flexible flatfoot were recruited to participate in the study. This study has been examined by the ethical review committee of Yueyang Hospital.

51 feet were treated with Kinesio Taping to relax plantar fascia, increase the longitudinal arches sensory input and correct the calcaneus valgus once a week and removed tapes after 3 to 4 days. At the same time, children walked on their heels and toes to exercise foot muscles every day. Gait analysis data and Foot Posture Index of 29 patients were collected before and six weeks after treatment. The common Helen Hayes model[3] was used for gait analysis. A 19-camera (Motion Analysis Corporation, Santa Rosa, CA) motion analysis system combined with 4 force plate (Bertec) was used to record kinematic and kinetic data. Footscan7 was used to record plantar pressure. The motion data and force plate data of five trials were analysed using motion software Orthotrack 6.6.1(Motion Analysis Corporation, Santa Rosa, CA). Five walking trails with ten to fifteen gait cycles were averaged for each participant.

Statistical analysis was performed using SPSS version 19(SPSS Inc., Chicago, USA). Using paired t-test and non-parametric Wilcoxon test to analyse the data.

RESULTS

After 6 weeks of treatment, the Arch Index and Foot Posture Index of flexible children declined significantly (AI z=-5.908, p=0.00; FPI z=-4.823, p=0.00) (Table 1). The Ankle Power and Ground Reaction Force (FZ1) increased significantly (AP t=-2.901 p=0.01; GRF1 z=-2.542, p=0.01) (Table 1). There were no significant differences in Knee Power, Hip Power before and after treatment (p>0.05). In terms of kinematic parameters, the ankle, knee and hip motion range in sagittal plane and the max flexion angle of knee in swing phase are no obvious change after treatment(p>0.05). The Gait Deviation Index increased, but the difference was not statistically significant (p>0.05) (Table 1).

Table 1: Gait function and foot posture of children with flexible flatfoot

	Before	**After six weeks**	***t/z***	***p***
Arch Index	29.9(28.8,31.3)	28.7(27.8,30.2)	-5.908	0.00
Foot Posture Index	8(6,10)	7(5,9)	-4.823	0.00
Ankle Power (W/kg)	1.64 ± 0.50	1.83 ± 0.46	-2.90	0.01
GRF1(N/kg)	1.08(1.03,1.14)	1.11(1.06,1.19)	-2.542	0.01
GDI	83.66(75.86,89.63)	84.71(78.85,89.40)	-0.900	0.37
Ankle ROM(°)	26.91±5.12	27.28±4.66	-0.66	0.51
Knee ROM(°)	61.49±6.71	61.89±5.40	-0.56	0.58
Hip ROM(°)	43.40±5.38	44.23±4.97	-1.54	0.13

Note: Joint power is standardized by dividing height and weight; GRF is standardized by weight and acceleration of gravity.

DISCUSSION

After six weeks of Kinesio Taping combined with muscle training, the height of the foot arch of children with flexible flatfoot was improved, while the ankle joint power and the ground reaction force were also improved. The walking function is better in dynamics, but not in kinematic parameters. In the short term, the improvement of longitudinal arch height may only improve the ankle power, but not affect the performance of the hip and knee joints power.

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DISCLOSURE STATEMENT

The authors declare no conflicts of interest.

A NEW METHOD TO EVALUATE THE DEFORMITY OF THE TALAR DOME

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INTRODUCTION

A flat-top talar dome is a deformity that can occur in patients after treatment of idiopathic clubfoot. Previous studies have noted variable rates of flat top talar dome deformity and its association with altered ankle mobility [1]. Manual measurements of plain radiographs have been used in studies that have quantified the deformity. The radius to length ratio (R/L ratio) of the talus is a common index of trochlear flattening determined by Mose rings [1-5]. The R/L ratio may incorrectly suggest or underestimate flattening based on talar length. In some cases, it does not fully describe the flattening deformity as it can be variable along the dome; previous studies have mentioned choosing one radius is difficult in these cases [1, 5]. This study describes a new image-processing algorithm to quantify the deformity of the talus following treatment of clubfoot.

CLINICAL SIGNIFICANCE

The flat-top talus deformity may impact daily activities and has been associated with increased risk for arthrosis [6]. This custom-written image processing tool may provide additional insight into the deformity that cannot be assessed using traditional radiographic measures.

METHODS

Skeletally mature patients previously treated for idiopathic clubfoot (ave. age 17.9±1.6 yrs) were identified from an IRB approved clubfoot registry. Participants without orthopedic conditions for whom foot radiographs were available were included as a control group. Weight bearing lateral foot radiographs were assessed in all participants. Manual radiographic measures of the talus included the radius of curvature (ROC), length, height, R/L ratio, alpha angle, radius of the tibial plafond, and the ratio of the radii of the talar dome and tibial plafond (R/R ratio) [1-3].

Using a custom image-processing MATLAB code, the talus was first segmented from the radiograph. Then, the end points of the articulating surface of the talus were selected manually. Three equally sized regions (anterior, central, and posterior) were then identified by the algorithm within the area bounded by the articulating surface. The average slope of each region of the talar dome from the posterior to anterior direction was estimated, and flatness was determined as the variance in the slopes across the three regions. Lower variance indicated a flatter dome.

In affected feet, inter-rater reliability (IRR) was determined for the MATLAB based (3 raters) and radiographic (4 raters) measures of talar dome flatness. Spearman's rho was used to determine correlations between traditional radiographic and MATLAB measures of flatness.

The lateral radiographs of the control group were examined by a clinician for exclusion of radiographs demonstrating false rotational profiles of normal talar domes (n=5 feet excluded) [7]. The average flatness of the talar dome of the clubfoot group, using both measurement methods, was compared to flatness of the control group using unpaired two-tailed student's t-tests.

As an example case, flatness measurements of the talar dome are presented using the radiographic technique and image processing technique in a skeletally mature patient previously treated for clubfoot, who had undergone 3D gait analysis as part of the registry.

DEMONSTRATION

Inter-rater reliability (IRR) for radiograph measures performed on 52 affected feet was excellent for the ROC of the talar dome (ICC=.985), talar length (ICC=.952), R/L ratio (ICC=.987), and alpha angle (ICC=.928). Radius of tibial plafond (ICC=.827) and talar height (ICC=.893) were highly reliable. The R/R ratio had moderate IRR (ICC=.608). The IRR for the MATLAB measure of talar dome flatness (n=15 affected feet) was very high (ICC=.895). Flatness of affected feet using MATLAB was strongly correlated with ROC of the talar dome (r=.621, p=.013), alpha angle (r=.557, p=.031), and R/R ratio (r=-.589, p=.021). Flatness using MATLAB was not correlated with R/L ratio which is the most common radiographic measure of deformity.

The average manual radiographic measures of control feet compared well to previously reported values (n=27 feet; R/L ratio= .364±.021) [5]. This R/L ratio was significantly different from that of affected feet (.561±.411; p =0.02). The average MATLAB flatness for the control and affected feet were .767±.156 and .372±.173, respectively, which were significantly different (p<0.001).

Lateral radiographs of the example case visually demonstrated unequal deformity (R>L) (Fig. 1). Upon examination of manual radiographic measures, the R/L ratios were near equal (R=.407, L=.408). Other measures were similar between sides, but the R/R ratio was higher for the left foot, indicating more congruency in articulation (R=.784, L=.957). The MATLAB measure of flatness was lower for the right foot, quantifying a flatter talar dome (R=.370, L=.599). Max ankle power for each side was similar (R= 3.38 W/kg; L= 3.42 W/kg). However, the right ankle had a decreased dynamic range of motion (R= 33°; L= 37°).

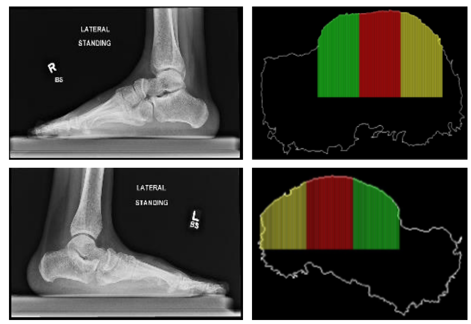

Figure 1: Standing lateral radiographs with MATLAB image processing visual output (right) of a participant previously treated for bilateral clubfoot

SUMMARY

There currently lacks a single measurement that fully describes the talar dome deformity. While manual radiographic measures can estimate the flatness of the talar dome, they have limitations and can fail to fully capture the morphology [1, 5]. The affected feet had higher variability in flatness measures using both methodologies, indicating the deformity is complex. In the case presented, the new method of flatness revealed a side to side difference that the traditional R/L ratio failed to show, and functionally, ankle mechanics reflected the differences in talar dome deformity between each side.

The new image processing technique to quantify the talar dome deformity was quick to apply (< 5 min per radiograph), had high IRR, and may provide additional insight regarding the morphology. Future studies should investigate the areas of increased flattening (i.e. anterior vs. posterior regions of the talar dome) and the relationship to ankle biomechanics.

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Quantifying pediatric gait from metrics derivable from wearable inertial sensors

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Introduction: Gait indices provide a valuable single score of gait function for clinical evaluation, screening, and outcome assessment for children with walking disorders, such as cerebral palsy (CP). However, indices widely used in clinical gait analysis require a laboratory setting. Use of lightweight wearable inertial sensors can bring accurate gait assessment to the orthopaedic clinic to quantify gait deviation from typically developing (TD) children. The aim of this initial study was to determine gait parameters that can be extracted from devices using inertial measurement units[1] (IMUs) that correlate to gait function, as assessed by the Gait Deviation Index (GDI), a lab-based 3D kinematic measure that provides a single score of gait function[2].

Clinical Significance: Establishing easily implemented gait assessment using lightweight wearable devices with inertial sensors can expedite diagnosis, treatment and outcome assessment.

Methods: Prospective analysis of 13 children with bilateral or unilateral CP and 8 TD children aged 7-11 (average 9.2±1.4) who consented and underwent 3D gait analysis from October 2016-February 2019 was conducted. Children with CP with Gross Motor Function Classification System (GMFCS) levels I-III and GDI $\leq$ 90 were included. As an initial step toward developing an IMU-based Pediatric Gait Index and building upon the previously developed Pediatric TDI[3], five gait parameters that can be extracted reliably from an IMU device were investigated for correlation with GDI, including: velocity, cadence, step length, single limb support (SLS) asymmetry, and trunk sway.

Results: SLS asymmetry and maximum trunk sway were most highly correlated to GDI (Figure 1.). Multivariate linear regression model using SLS asymmetry and maximum trunk sway explained 64% of GDI variance (R^2=0.64). Multivariate logistic regression model using SLS asymmetry and maximum trunk sway classified children scoring 2 SD below mean GDI with 78% sensitivity, 83% specificity.

Discussion: Clinical gait assessment based on SLS asymmetry and maximum trunk sway may offer a revealing gait evaluation that is readily conducted in the clinic, using lightweight wearable IMU-based devices. Further research with a larger sample can determine a single score IMU-based gait index to assess walking abnormalities in the clinic. Detecting gait abnormalities in the clinic can accelerate diagnosis, treatment, and outcome assessment.

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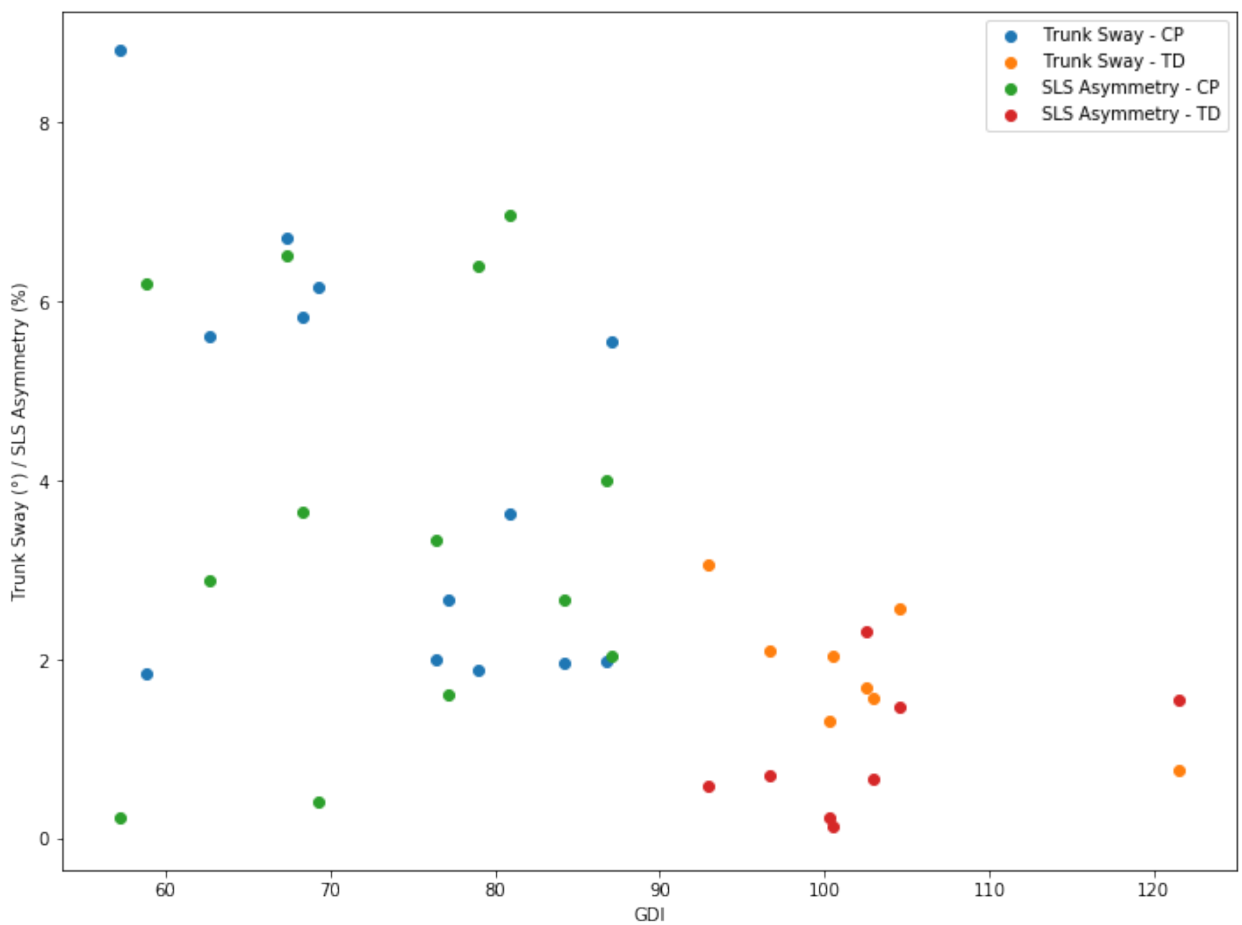
Figure 1. Maximum trunk sway and single-limb support in relation to the Gait Deviation Index (GDI) in children with cerebral palsy and typically developing (TD) children.



VARIABILITY IN THE EXPRESSION OF CRAWLING LOCOMOTION IN THE NEWBORN RAT: CHARACTERIZATION OF STEPPING PATTERNS AND INTERLIMB COORDINATION

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INTRODUCTION

The study of the expression of locomotion in human infants has helped understand the development of important skills such as crawling and walking. Studies in newborn human infants are often limited to describing the presence of movement and not to the evaluation and description of how movement is expressed at such an early age. Animal studies with newborn rats provide the opportunity of understanding the neural and behavioral development of movement and the emergence of locomotion during early development. In rats, crawling, or an early form of walking, it is not distinctively expressed until day 8 after birth. However, when provided with a stimulus of ecological relevance and placed on an appropriate surface, crawling can be expressed on day 1 after birth. Studies from our laboratory showed that crawling can be evoked by exposure to amniotic fluid (AF) or milk 24 hours after birth. Although this expression of crawling looks organized and resembles mature crawling, further studies showed that when studied closely, crawling at that early age lacks the behavioral organization expressed in a typical diagonal-sequence-diagonal-couplet pattern. These findings challenge studies conducted *in vitro* that suggest that newborn rats are physiologically prepared to exhibit a coordinated pattern. In this presentation, the expression of crawling in the 1-day-old rat will be discussed by characterizing the stepping pattern using the paradigm of AF and milk exposure to evoke crawling.

CLINICAL SIGNIFICANCE

Understanding the variability of gait, stepping patterns, and coordination in early development helps establish the parameters needed for understanding adult locomotion. As this study shows, having the physiological capacity it is not enough as other aspects will play a role in the expression of movement and locomotion. Some of these factors include the surface in which an individual locomote, the ratio of muscle and mass, and other constraints such as the clothes or diaper an infant or adult is wearing. This study emphasizes the importance of understanding locomotion as a process of active learning and adaptation to the physical characteristics of the body and the environment in which the individual is moving.

METHODS

One-day-old rat pups were exposed to 0.3 mL of AF collected on day 20 of gestation or commercially available half-and-half milk. Fluids were presented inside a 1.5 mL microcentrifuge tube that was placed over the pup's snout during testing. All pups expressed crawling during odor presentation. Video recordings from a ventral camera view (through glass) were used to observe the crawling response and to characterize patterns of stepping. Behavioral analysis included the characterization of the sequence of steps (footfalls), interlimb phase, the timing of steps, and step cycle (stance and swing phases).

RESULTS

Data analysis included the quantification of the onset and offset of steps performed by each of the limbs (i.e., right and left hindlimbs and right and left forelimbs). In addition, scoring included quantifying any behavior that represented a disruption of the stepping cycle, including rotation of the torso, flipping onto their backs, or losing contact with the testing tube containing the solution. Results showed that the statistical distribution of interlimb phases was very variable. While the expected relative phase in an adult is of .5 in relation to a reference limb, newborn pups are expressing highly variable relative phases. Statistical analysis showed that phase relationships between limbs were expressed at random, and their gate was not the typical diagonal-sequence-diagonal-couplet pattern.

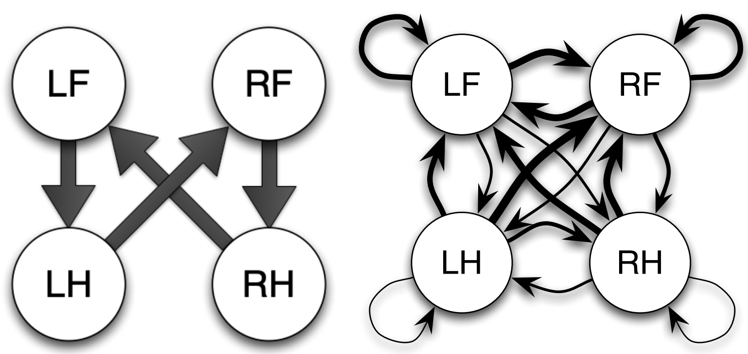

Figure 1. The left figure shows a typical Diagonal Sequence Diagonal Couplet walking gait (characteristic of most mammals). The right figure shows the step sequences of the newborn rat. Circle represents footfalls (onset of stance). The arrows show sequencing and width is proportional to the sequencing transition (how likely is one foot to follow another).

DISCUSSION

Our results indicate that although newborn rats can express crawling, its expression does not resemble the depiction of a coordinated stepping pattern reported in studies of stepping on treadmills, air-stepping, or spinal cord explants *in vitro*.

DISCLOSURE STATEMENT

V. Mendez-Gallardo & S. Robinson have no conflicts to disclose.

GAIT ANALYSIS OF LEPROSY PATIENTS WITH FOOT DROP

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Introduction

World Health Organization registered, in 2018, 208.619 new cases of leprosy globally. Most of the cases occurred in India, being Brazil the second. In leprosy patients (LP), the Mycobacterium leprae affects peripheral nerves, such as the tibial and the common peroneal. This condition results in permanent loss of the protective sensibility of the sole and dorsiflexion weakness, as well as equinovarus deformity [1] and walking impairment [2]. Drop foot deformity is a common problem. The propose of this study is to describe, using 3D motion analysis, the gait characteristics of LPs with common peroneal nerve damage and drop foot. In our knowledge, this is the first work using 3D motion analysis to describe leprosy patients' gait.

Clinical significance

Describing and understanding the dynamics of gait in leprosy subjects may provide further insight into how the foot drop affects walking movement, helping clinicians to improve the rehabilitation programs and surgical approaches, and their follow-up.

Methods

Twelve Leprosy patients (LP) (8 women; 44.7±10.8 years; 82.6±20kg; 1.7±8.2m) with foot drop and fifteen control subjects (CG) (3 women; 37.6±15.6 years; 74.5±18.6kg; 1.7±6.1m) were evaluated. LP subjects were cured by multidrug therapy. The study was approved by the Ethics Committee of the UFRJ University Hospital. They were required to walk barefoot at their self-selected gait speed on a 10 m walkway with four embedded force plates BTS P-6000 (BTS, Italy) sampling at 400 Hz. Kinematic were collected by eight infrared cameras (SMART-D, BTS, Italy) sampling at 250 Hz, using 18 markers placed according to the Helen Hayes protocol. Dynamic gait and force data were reconstructed using the Smart Analyzer Software (BTS, Italy). Averaged data from five trials were obtained for kinematics and ground reaction data of the right lower limb from CG, and affected (AL) and unaffected limbs (UL) of LP. After checking for normality using Shapiro-Wiki test, data were compared between AL and CG, and between UL and CG using independent t-test. AL and UL data were compared using paired t-test. The significance level was $\alpha = 0.05$. The statistical tests were performed in SPSS 17.0 (SPSS, USA).

Results

LP walked with lower gait speed (0.8 ± 0.2m/s) compared with CG (1.1 ± 0.2m/s) ($p=0.003$). LP walked with higher plantarflexion at heel strike and lowered plantarflexion at push-off in the AL compared to UL and CG (Table 1). Moreover, during balance, dorsiflexion was significantly lesser in AL compared to UL and CG. LP also presented higher knee adduction-abduction and pelvis tilt range compared to CG. Ground reaction differences were observed in the force during the push-off, with lower values at AL compared to UL and CG. Additionally, plantarflexion moment from AL was lower compared to CG (Figure 1A), and the hip flexion moment from LP (both sides) was lower compared to CG (Figure 1C). LP subjects walked with lower ankle total work at stance in the AL (-0.03±5.4 J/Kg) compared to controls (-4.6±5.2 J/Kg) ($p=0.04$) and higher ankle total work at balance (0.06±0.03 J/Kg) compared with CG (-0.01 ± 0.08 J/Kg) ($p<0.001$).

Table 1 – Gait parameters from LP and CG

	Leprosy subjects AL	Leprosy subjects UL	CG	*p* (AL – UL)	*p* (AL - CG)	*p* (UL - CG)
Ankle Initial contact (*deg*)	-16.8 ± 8.3	-6.6 ± 10.3	-5.4 ± 2.5	**0.01***	**0.02***	0.704
Plantarflexion push off (*deg*)	-29.1 ±11.5	-14.6 ± 11.6	-18.8 ± 5.8	**0.018***	**0.023***	0.286
Peak dorsiflexion balance (*deg*)	-12.4 ±6.2	2.4 ± 7.6	-1.4 ± 3.9	**0.001***	**<0.001***	0.115
Peak knee flexion stance (*deg*)	15.5 ± 10.0	17.4 ± 8.6	13.3 ± 4.6	0.534	0.497	0.125
Peak knee flexion balance (*deg*)	55.9 ± 17.4	51.8 ± 12.8	57.2 ± 11.2	0.533	0.959	0.237
Knee range adduction stance (*deg*)	12.6 ± 9.3	10.6 ± 5.5	6.1± 2.9	0.476	**0.046***	**0.047***
Peak hip flexion stance (*deg*)	32.1 ± 10.8	34.2 ± 10.8	32.3 ± 7.3	0.279	0.939	0.605
Peak hip extension stance (*deg*)	-5.9 ± 11.3	-6.7 ± 10.0	-6.0 ± 6.4	0.649	0.964	0.835
Peak hip flexion balance (*deg*)	39.4 ± 11.8	36.4 ± 11.0	34.8 ± 6.7	0.248	0.180	0.799
Plevic tilt range (*deg*)	5.1 ± 2.4	4.6 ± 1.7	3.1 ± 0.9	0.474	**0.030***	**0.009***
Pelvic obliquity range (*deg*)	6.9 ± 3.0	6.6 ± 2.8	8.1 ± 5.0	0.097	0.440	0.313
Pelvic rotation range (*deg*)	12.0 ± 4.5	11.0 ± 4.7	11.4 ± 3.9	0.201	0.686	0.721
First peak of force *(%BW)*	101.5 ± 10.9	98.6 ± 5.4	97.8 ± 4.0	0.373	0.073	0.149
Second peak of force *(%BW)*	98.6 ± 5.2	100.7 ± 8.7	104.1 ± 5.5	0.534	**0.018***	0.241

AL – Affected limb; UL – Unaffected limb. * Bolded values indicate significant differences

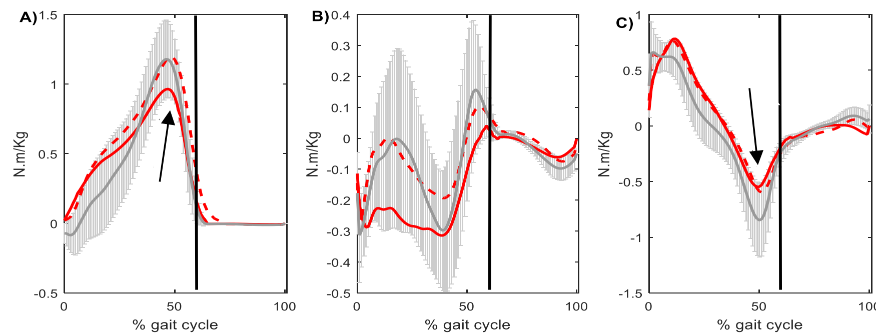

Figure 1 – Ensemble averages of sagittal ankle (A), knee (B), and Hip (C) moments during gait for the control group (grey line), AL (continuous red line), and UL (dotted red line) from leprosy patients. Shaded areas represent 1 standard deviation above and below the control group mean. The arrows indicate the statistical difference between groups and the black vertical line the toe-off.

Discussion

The walking ability was affected in LP subjects with foot drop, as observed in [2]. The LP patients presented difficulties for clearing the floor at heel strike and in the swing phase of gait, caused by impaired dorsiflexors. During stance, lower ankle range of motion and torque were observed, which can be associated with reduced gait speed. Reduced plantarflexion moment may reflect a possible weakness of the gastrocnemius-soleus complex due to tibial nerve involvement, causing disruption in the terminal coupling mechanism between the ankle and the knee [3]. Achilles tendon is frequently lengthened during the surgical procedure for the correction of drop foot, which may adversely affect the plantar flexion power to an already compromised gastro-soleus unit.

Acknowledgments

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Disclosure statement

The authors declared no potential conflicts of interests with respect to this article

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THE INFLUENCE OF DIFFERENT MUSCULOSKELETAL MODELS FOR ESTIMATED KNEE CONTACT FORCE DURING GAIT AFTER TOTAL KNEE REPLACEMENT USING OPENSIM SIMULATION

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INTRODUCTION

Musculoskeletal modeling is commonly used to estimate the underlying muscle activations and joint contact force to better understanding human movement. For example, the knee contact force (KCF) of patients after total knee arthroplasty (TKA) are of great value to implant designers to estimate the wear rate and stress distribution in the implant. However, the effect of musculoskeletal model choice on estimates of KCF during gait is largely unknown. The purpose of this study was to evaluate different musculoskeletal models, focusing on how the number of muscles and DOF of knee joint influence the estimated KCF during gait.

CLINICAL SIGNIFICANCE

The knowledge gained could benefit clinicians and implant designers by providing a better understanding of the influence on different numbers of muscles and DOF of the knee joint to the predicted KCF when using musculoskeletal modeling.

METHODS

The experimental data were taken from the second through fifth “Grand Challenge Competitions to Predict In Vivo Knee Loads” [1]. Briefly, four subjects with unilateral knee instrumented implants were included in this study. For each subject, three over ground trials were recorded under the normal gait pattern at a self-selected speed. Data collected included maker trajectories, ground reaction forces, EMG and in vivo knee loading, which was calculated from the load cells embedded in the tibial tray using a validated regression equation [2].

Gait2354, Gait2392, Gait2354 and Gait2392 with 3 degrees of freedom (DOF) of knee joint were used in this study. All four models consist of 12 rigid body segments and the same DOFs for all joints except the knee. Originally, a 1 DOF knee joint with proximodistal and anteroposterior translations as a function of knee flexion were used. Two additional DOF (ab/adduction and rotation) were developed in Gait2354 and Gait2392 with 3 DOFs of the knee joint.

OpenSim 4.0 was used to generate the muscle-driven gait simulation. The same settings were firstly used in the scale and inverse kinematics steps for each of the four musculoskeletal models. Then, residual reduction and computed muscle control algorithms were used to generate muscle activations and forces to drive the dynamic gait simulation. Finally, the KCF was determined by the joint reaction program [3].

The correlations of EMG data and normalized muscle force were determined with Pearson correlation. One-way repeated ANOVA was used to assess differences in peak KCF among the four models and in vivo measurements. Post-hoc tests were performed for multiple comparisons. The results were considered statistically significant when $P < 0.05$.

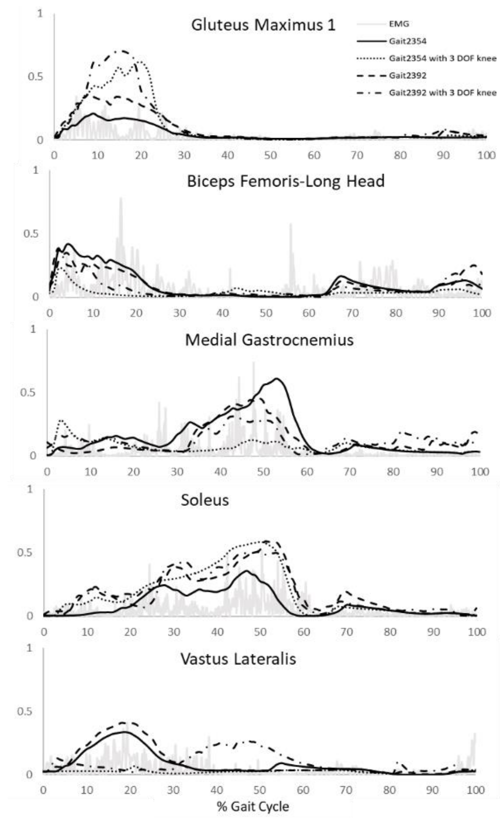

Fig.1 Comparison of EMG data for one subject with muscle force predicted by four models

RESULTS

For Gait2392, significant correlations between muscle forces and EMG data were found for seven muscles, including gluteus maximus, biceps femoris, semimembranosus, vastus medialis, vastus lateralis, medial gastrocnemius and soleus. For Gait2354, the correlations were significant for six muscles, including gluteus maximus, adductor magnus, biceps femoris, vastus lateralis, medial gastrocnemius and tibialis anterior. For Gait2354 and Gait2392 with 3 DOFs of knee joint, the muscle forces of gluteus maximus, soleus showed a significant correlation with EMG data (Fig.1). The four musculoskeletal models generally overestimated the peak KCFs compared to the in vivo measurement, especially for Gait2392 with 3 DOFs of knee joint. Gait2392 has the smallest difference between model prediction and in vivo measurement for both peak KCF (Table 1).

Table 1 Comparison peak KCFs between in vivo measurement and predicted values

Model	Subject	First peak of KCF (BW)			Second peak of KCF (BW)		
		Predicted	In vivo	Difference	Predicted	In vivo	Difference
Gait2354	Mean (SD)	3.83(0.41)*	2.28(0.09)	1.55(0.66)	5.61(0.53)	2.42(0.15)	3.18(1.09)
Gait2354_3 DOFs knee	Mean (SD)	4.47(0.15)#	2.28(0.09)	2.18(0.37)	5.63(1.15)	2.42(0.15)	3.20(0.2.17)
Gait2392	Mean (SD)	3.44(0.29)*	2.28(0.09)	1.16(0.41)	3.78(0.27)*	2.42(0.15)	1.36(0.56)
Gait2392_3 DOFs knee	Mean (SD)	6.77(0.55)#	2.28(0.09)	4.49(0.92)	5.56(0.35)#	2.42(0.15)	3.14(0.56)

DISCUSSION

The results suggest that increasing the number of muscles could improve the estimated muscle force and peak KCF in some extent, but adding additional two DOFs of knee joint in the coronal and frontal planes without ligament structures worse the predicted KCF results.

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DISCLOSURE STATEMENT

There are no conflicts of interest to disclose.

Pelvic Tilt in Patients with Acetabular Dysplasia during Functional Gait Tasks

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Introduction

Our study was intended to evaluate the change in pelvic tilt of patients with hip dysplasia during gait. Prior studies of gait mechanics in hip dysplasia patients have demonstrated reduced hip joint forces, lateralization of the hip joint center, reduced hip flexion, reduced hip adduction, and mixed findings regarding lower peak hip extension.[1-4] Harris et al. suggested that pelvic tilt in hip dysplasia patients had little variance with no clear minimum or maximum value during gait. We hypothesized that patients with hip dysplasia exhibit compensatory change in anterior pelvic tilt for better coverage of the femoral head along the load bearing axis.

Clinical Significance

Describing baseline gait dynamics in patients with acetabular dysplasia provides a marker for change following clinical or surgical intervention.

Methods

Ten consecutive patients were studied prior to undergoing periacetabular osteotomy for symptomatic acetabular dysplasia. Walking gait was analyzed using motion capture technology (Vicon, Oxford, UK). The motion capture data were analyzed using Visual3D software (C-motion, Germantown, MD.). Each patient was recorded while walking on level, 10° incline, and 10° decline surfaces. Pelvic tilt change during gait was measured relative to each patient's stationary weight-bearing pelvic tilt position. Data were recorded over one minute of continuous walking and averaged. Standing pelvic tilt was evaluated using the pubic symphysis to sacroiliac (PS-SI) joint measurement on upright anteroposterior radiographs. Local weighted regression with Loess fitting was used to estimate the relationship between the mean pelvic tilt (◦) and gait cycle (%) on level, incline, and decline surfaces. Spearman correlation coefficient was estimated between the average pelvic tilt and PS-SI distance.

Results

On level surface, 5 patients exhibited an anterior pelvic tilt change and the remaining 5 had posterior pelvic tilt change. The average mean, maximum, and minimum pelvic tilt change was -0.88°, 0.88°, and -2.73°, respectively. On the incline surface, all patients had posterior pelvic tilt change. The average mean, maximum, and minimum pelvic tilt change was -4.34°, -2.48°, and -5.93°, respectively. On the decline surface, 5 patients had an average anterior pelvic tilt change and the remaining 5 patients had and average posterior pelvic tilt change. The average mean, maximum, and minimum pelvic tilt change was -0.08°, 1.38°, and -1.56°, respectively. The average PS-SI measurement was 88.55mm ± 14.88. Average pelvic tilt change was negatively correlated with PS-SI distance on all 3 surfaces using a Spearman coefficient (Table 1).

Table 1. Spearman rho between average pelvic tilt and PS-SI distance by surface

Surface	Spearman rho	p-value	95% CI
Level	-0.370	0.293	-0.811 to 0.339
Incline	-0.358	0.310	-0.805 to 0.351
Decline	-0.103	0.777	-0.688 to 0.563

Discussion

Our results demonstrate that, on average, patients with hip dysplasia exhibit variable change in pelvic tilt while walking on different surface types. A weak negative correlation with standing pelvic tilt suggests that those with less anterior pelvic tilt while standing rotate anteriorly during gait, while those with more anterior standing pelvic tilt rotate posteriorly during gait. Dynamic insights from this pilot prospective study expand our understanding of patho-mechanics during gait in acetabular dysplasia, which may guide future comparisons with pre- and post-surgical states and functionality in such patients.

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Disclosure Statement

The authors have nothing to disclose.

Pelvic Tilt in Patients with Femoroacetabular Impingement during Functional Gait Tasks

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Introduction

Our research was designed to evaluate pelvic tilt during ambulation in patients with femoroacetabular impingement (FAI). Prior studies have shown paradoxical anterior pelvic tilt in FAI patients.[1,2] We hypothesized that patients with FAI have compensatory posterior pelvic tilt while walking to relieve anterolateral hip impingement.

Clinical Significance

Describing baseline gait dynamics in patients with FAI provides a marker for change following clinical or surgical intervention.

Methods

5 women and 3 men with FAI were analyzed prior to surgical intervention for FAI, using motion capture while ambulating (Vicon, Oxford, UK). The motion capture data was analyzed using Visual3D software (C-motion, Germantown, MD). Patients were recorded while walking on a flat, 10° incline, and 10° decline surface. The pelvic tilt was measured relative to each patient's stationary weight bearing pelvic tilt. Data were recorded over one minute of continuous walking and averaged together. Pelvic tilt was also evaluated using the pubic symphysis to sacroiliac joint measurement (PS-SI) on standing anteroposterior radiographs. Local weighted regression with Loess fitting was used to estimate the relationship between the mean pelvic tilt (◦) and gait cycle (%) on level, incline, and decline surfaces. Spearman correlation coefficient was estimated between the average pelvic tilt and PS-SI distance.

Results

On level surface, 7 patients had an average anterior pelvic tilt and 1 had posterior pelvic tilt. The average mean, maximum, and minimum pelvic tilt was 2.72°, 4.26°, and 1.15°, respectively. On incline surface, 7 patients had average posterior pelvic tilt and 1 patient had average anterior pelvic tilt. The average mean, maximum, and minimum pelvic tilt was -1.98°, –0.48°, and -3.49°, respectively. On decline surface, all patients had an average posterior pelvic tilt. The average mean, maximum, and minimum pelvic tilt was -4.56°, -3.19°, and -6.22°, respectively. The average PS-SI measurement was 98.03mm +/- 18.07. Average pelvic tilt was negatively correlated with PS-SI distance on level and decline surfaces and was positively correlated on incline surface using the Spearman coefficient (Table 1).

Table 1. Spearman rho between average pelvic tilt and PS-SI distance by surface

Surface	Spearman Rho	p-value	95% CI
Level	-0.333	0.4198	-0.841 to 0.485
Incline	0.0238	0.9554	-0.692 to 0.716
Decline	-0.167	0.6932	-0.780 to .0.610

Discussion

Our data demonstrate that, on average, patients with FAI adopt anterior pelvic tilt while walking on level surfaces and posterior pelvic tilt while walking on incline and decline surfaces. Standing pelvic tilt was weakly negatively correlated with average pelvic tilt during gait on two of three surface grades, suggesting that patients with less anterior pelvic tilt while standing anteriorly rotate during gait, while patients with more anterior standing pelvic tilt posteriorly rotate during gait. This data may suggest a regression to the mean while walking. The combination of FAI morphology with anterior pelvic tilt during gait may explain why some patients present with symptomatic FAI while others remain asymptomatic.

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Disclosure Statement

The authors have nothing to disclose.

GAIT AND RANGE OF MOTION ANALYSIS IN HIP DYSPLASIA AND FEMOROACETABULAR IMPINGEMENT: DISTINGUISHING HIP PATHOLOGY WITH HIP PATHOMECHANICS

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INTRODUCTION

Femoroacetabular impingement and hip acetabular dysplasia account for the majority of hip disease. However, current research investigating the complex relationship between hip pathology and range of motion and provocative gait tests has not taken into account gait and range of motion analysis.

Hip dysplasia and femoroacetabular impingement should be linked to gait disturbance because the aberrant arthrokinematics associated with the pathologies produce early joint degeneration, which generates pain and is followed by mechanical compensation when moving.

CLINICAL SIGNIFICANCE

This study compared gait and range of motion differences in patients with hip dysplasia and FAI to assess if the presence of a particular hip disease could be evaluated by gait analysis. The ability to distinguish these two prevalent hip pathologies purely based on pathomechanics holds immense diagnostic and clinical benefits.

METHODS

A prospective, non-randomized cohort study was conducted with 10 cases of hip dysplasia and 10 cases of FAI. The gait and range of motion tests were indicated for patients whose histories, clinical measurements, and radiographic findings confirmed either hip dysplasia or FAI.

An eight-camera motion capture system (Motion Analysis Corp.) collected kinematic data pre-surgery and biomechanical analysis software (Visual3D) was used to model joint kinematics.

Sagittal peak flexion, sagittal peak extension, frontal range of motion, and transverse range of motion at level, incline, and decline slopes were statistically analyzed with t-tests while controlling for age and BMI.

RESULTS

Patients with hip dysplasia exhibited significantly increased level flexion (p=0.033), incline extension (p=0.052), decline flexion (p=0.052), and decline extension (p=0.043) when compared to patients with FAI. A total of 20 patients (17 female and 3 male) with a mean age of 29.1 ± 8.5 years and BMI of 26.2 ± 4.7 years were included in the current study. There were 9 left hips and 11 right hips.

DISCUSSION

Hip dysplasia patients display greater flexion on level walking and extension on incline walking than FAI. The differing gait presentations between dysplasia and FAI likely arise from their unique morphological deformities and may aid in diagnosis and management of these complex disorders.

DISCLOSURE STATEMENT

Both Jason Lin and Dr. Joel Wells have no conflicts of interest to disclose.

The interaction of assistive devices and propulsive forces during walking post-stroke

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INTRODUCTION

Stroke is the primary cause of long-term adult disability in the United States and more than 40% of stroke survivors have moderate or greater functional impairments [1]. Individuals post-stroke frequently have decreased walking speed, which is correlated with a reduction in paretic (weak side) propulsive force. Thus, increasing paretic propulsion is an important goal of rehabilitation [2]. Assistive devices, such as treadmill handrails or a cane, are commonly utilized during rehabilitation. Research has demonstrated that these devices alter biomechanics; step symmetry [3] and energy cost [4] of walking change with the use of handrails, and a cane provides braking assistance [5]. However, the direct impact of assistive devices on propulsive forces is unknown. Additionally, how the interaction between user and device may vary with differing levels of functional ability is not known. Therefore, the purpose of this study is to determine the effect of assistive device use, such as cane or handrails, on propulsive forces in individuals with varying functional level and experience with assistive devices. We hypothesized that the relationship between assistive device use and propulsive force will vary as a function of a subject’s level of functional impairment. We further hypothesize that peak assistive device propulsive forces generated on a handrail will be greater than on a cane.

CLINICAL SIGNIFICANCE

The results from this study will help elucidate the interaction between assistive device use and walking function post-stroke, to provide insight on how best to use assistive devices during rehabilitation and to maximize walking functional recovery.

METHODS

Three participants (2M, 1F; age 67 ± 11.36 yrs; 1.77 ± 1.60 yrs since stroke) from an ongoing study with different levels of function: no history of assistive device use, past history of an assistive device use, and current dependence on an assistive device, were included in this analysis. Inclusion criteria included: age 19-80, single-chronic stroke, and ambulatory. Motion capture analysis using a full-body, 65-marker set was done both overground, using in-ground force plates and an instrumented cane, and with an instrumented treadmill and handrails. For the treadmill trials, subjects walked for three walking conditions (3 minutes each): no handrail use, light support handrail use and self-selected handrail device use. Real-time feedback of the handrail forces was displayed for the light-support handrail use condition, to ensure participants were applying <5% of their body weight onto the handrails. For the overground trials, subjects completed three conditions: walking with no cane, walking with a cane using light pressure and walking with a cane comfortably. Real-time feedback of the cane forces was displayed for the light-support cane use condition on a monitor in front of the walkway, to ensure the force on the cane was below 5% of their body weight. One subject, who was dependent on an assistive device, was only able to complete the handrail and cane conditions. The peak paretic propulsive GRF and peak assistive device propulsive force were calculated for each subject and for each trial.

RESULTS

The peak propulsion did not differ greatly between treadmill conditions, but showed more variance overground (Table 1), particularly for subjects who did not require an assistive device to ambulate. Overall, the peak propulsion was 116-231% greater for the overground conditions compared to the treadmill. For both assistive device conditions, 108-395% greater peak assistive device anterior force was used in the self-selected compared to the light force condition. Additionally, 87-265% more anterior force was generated on the handrails compared to the cane.

Table 1: Peak paretic propulsion force and peak assistive device propulsion (%BW)

		Treadmill			Overground		
	Level of Function	**No HR**	**<5% HR**	**SS HR**	**No Cane**	**<5% Cane**	**SS Cane**
Peak Propulsive GRF (%BW)	No Assistive Device	6.9%	7.7%	7.9%	12.9%	9.5%	10.5%
	Past Assistive Device	6.6%	6.3%	5.9%	11.6%	9.2%	6.8%
	Device Dependent	N/A	2.4%	2.1%	N/A	4.1%	4.7%
Peak Anterior Assistive Device Force (%BW)	No Assistive Device	0%	1.0%	1.6%	0%	0.6%	0.6%
	Past Assistive Device	0%	0.7%	1.0%	0%	0.3%	1.1%
	Device Dependent	N/A	0.6%	1.3%	N/A	0.3%	0.5%

DISCUSSION

The purpose of this study is to determine the effect of assistive device use on propulsion in individuals post-stroke. Interestingly, there is little variation in peak propulsion across treadmill conditions. In contrast, participants who were not device-dependent saw a decrease in peak propulsion with cane usage greater than the minimal detectable change of 0.8% BW [6] (Fig. 1). These preliminary results suggest that while handrail use might not alter peak propulsion, using a cane can have a negative effect on these forces. As mentioned above, an individual's functional level altered how their propulsion changed between conditions, suggesting that functional level plays a role in the impact of assistive devices on propulsion. Continued work in this study will include additional subjects and regression modeling based on walking speed and functional level.

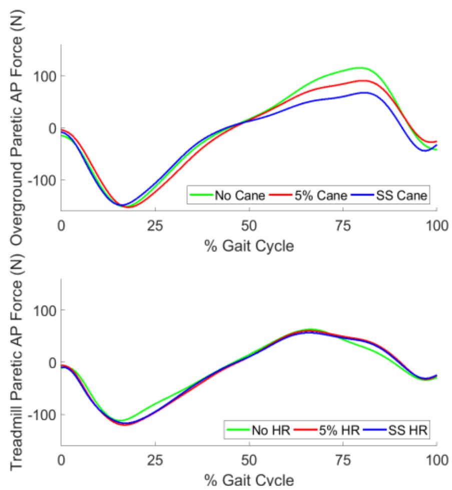

Figure 1. Paretic AGRF for a participant with previous assistive device use.

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DISCLOSURE STATEMENT

We have no conflicts of interest to disclose.

DIFFERENCES IN OVER GROUND WALKING AND STAIR CLIMBING BETWEEN ANKLE ARTHROPLASTY CANDIDATES AND CONTROLS

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INTRODUCTION

Over the last two decades, total ankle arthroplasty (TAA) has become the increasingly popular surgical treatment of end-stage arthritis of the ankle joint compared to ankle arthrodesis (AA) [1]. Several studies have been conducted to examine the efficacy of TAA versus the traditional surgical solution of AA, with the most recent study showing that both TAA and AA are effective at two-years after implementation, but the TAA has more improved patient reported outcomes [1, 2]. However, little has been done to examine how TAA affects the motion of the foot. One examined gait between TAA patients and controls modeling the foot as a single segment and only reporting spatio-temporal parameters and changes in ground reaction force patterns [1].

The study presented here will utilize the validated four segment (lower leg/tibia, hindfoot, forefoot, hallux) Oxford foot model [3] to examine the differences in the foot kinematics in a patient before TAA compared to a group of healthy ankles. This lays the ground work to examine how end-stage arthritis of the ankle changes the motion within the foot, as well as how treatments for end-stage arthritis (TAA or AA) then alters the motion within the foot.

CLINICAL SIGNIFICANCE

Ankle replacements aim to return patients to full function, though we know that they reduce the range of motion attained at the ankle joint. The insight gained through this study will better inform total ankle implant designs, as well as illuminate how the range of motion at the ankle changes with TAA versus AA.

METHODS

Full-body experimental motion capture data was collected for 4 controls (3M,1F; 30 ± 9.1 y.o.) and 4 ankle replacement candidates (4M; 63 ± 4.7 y.o.), using Vicon Nexus (Oxford Metrics, Oxford, UK) and a full-body marker set which included markers necessary for the Oxford foot model [3]. Subjects performed two activities for this study: (1) Walked at self-selected speed over level ground (15 m); (2) Climbed an instrumented stair case consisting of 3 stairs. Kinematic data were then exported to Matlab (MathWorks, Natick, MA, USA) for analysis. The stance phase was determined to be heel strike to toe off for the given limb, for both walking and stair climbing. Spatio-temporal parameters (including stride length, velocity, percentage of whole gait cycle where toe-off occurs, and percentage of gait cycle spent in single and double support) were calculated for both activities. Foot kinematics were determined using the Oxford foot model to find the angles between the hindfoot and tibia (HFTB), the forefoot and tibia (FFTB), the forefoot and hindfoot (FFHF) in the three anatomical planes. All angles are reported as the deviation from the kinematics collected during the static trial, where patients stood upright on both feet [3].

RESULTS

The candidates for TAA are showing signs of difficulty walking in their spatio-temporal gait parameters. Specifically, compared to the control group, the patient is taking shorter strides (0.98 m vs 1.38 m), walking slower (0.78 vs 1.25 m/s), delaying their toe raise to later in the gait cycle (67% vs 62%) and spending more time in double support (35% vs 23% of gait cycle). When climbing stairs, patients are also moving slower than the control group (0.41 vs 0.65 m/s). All results significant at $p<0.01$.

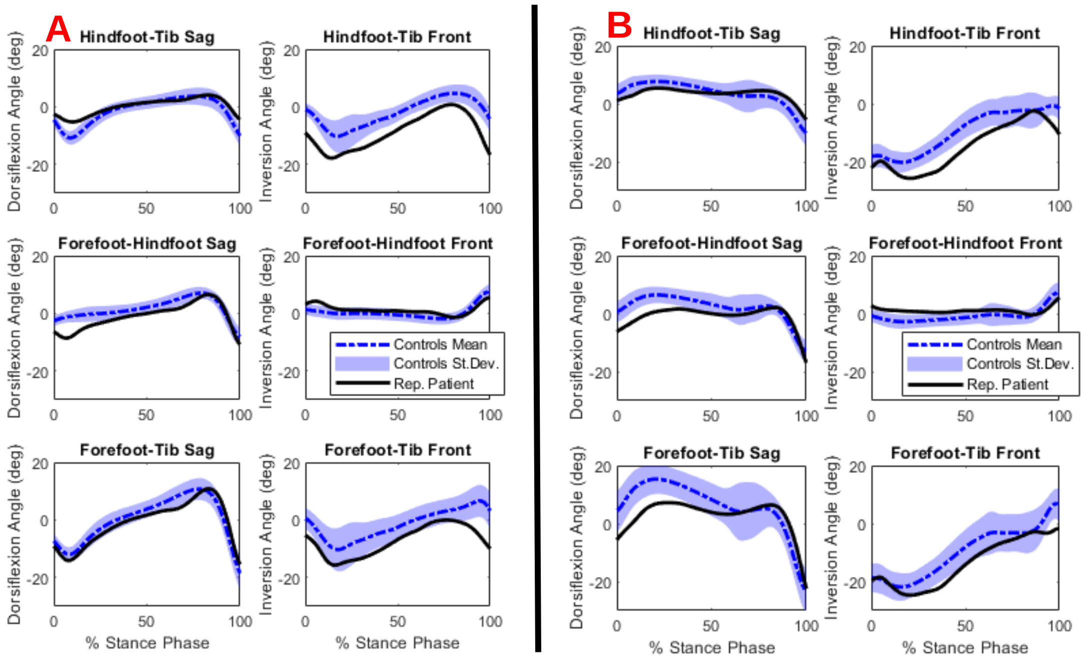

Figure 1. Foot segment kinematics for representative patient vs. control group. (**A**) Walking on level ground. (**B**) Stair climbing. Sagittal kinematics use dorsiflexion as positive. Frontal kinematics use inversion as positive.

When examining the foot kinematics, differences between the representative patient and the control group can be seen in the hindfoot-tibia joint and the forefoot-hindfoot joint, in both the sagittal and frontal planes, for both walking over level ground and climbing stairs (*Figure 1*). These differences do not appear as much in the forefoot-tibia angle calculation.

DISCUSSION

In the first rocker while walking over level ground, the control group plantarflexes in their hindfoot, with little motion between the forefoot and the hindfoot. The TAA patient shown has increased motion between their forefoot and hindfoot, which results in their whole foot plantarflexing due to the forefoot plantarflexing relative to the hindfoot in addition to the small plantarflexion of the hindfoot.

When climbing stairs, the patient is able to obtain similar hindfoot motion to the controls. However, the patient is unable to attain the doriflexed position of the forefoot due to the increased angle between the forefoot and the hindfoot. The inability to reach necessary dorsiflexion for climbing stairs is a common complaint heard by physicians from patients with severe ankle osteoarthritis.

This method clearly shows that differences exist between these populations. This work enables researchers to use this method to track these patients through recovery from TAA to determine if their foot kinematics get closer to the healthy population, as well as how kinematics differ between TAA and AA recovery.

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DISCLOSURES

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Post-Operative Outcomes of Pediatric Patients with Syndromic Hip Dysplasia

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INTRODUCTION

Hip dysplasia is a commonly experienced orthopedic problem for patients with a wide range of syndromic disorders and chromosomal abnormalities. Pediatric hip dysplasia can lead to long term complications in adulthood such as dysfunction, pain, and osteoarthritis if not properly managed [1]. Surgical treatment of hip dysplasia for skeletally immature patients is currently modeled after cerebral palsy (CP) treatment principles and can include a varus derotational osteotomy (VDRO) as well as a Dega-type pelvic osteotomy (PO). While outcomes of these procedures have been studied in cohorts with a unifying diagnosis, it has not been studied with a broader scope. This retrospective study aimed to assess the success of reconstructive surgery in the treatment of hip dysplasia in pediatric patients with a known diagnosis of a syndrome or chromosomal abnormality who underwent VDRO and/or Dega PO.

CLINICAL SIGNIFICANCE

Surgical intervention is often necessary to maintain ambulation, reduce pain, and decrease long term morbidity for patients with hip dysplasia. Further understanding of the pathology of syndromic hip dysplasia is necessary to identify more reliable reconstructive techniques to address hip instability.

METHODS

Medical charts of pediatric patients with a known diagnosis of a syndromic disorder or chromosomal abnormality who underwent reconstructive hip surgery for hip instability were reviewed. AP Pelvis x-rays were measured pre- and post-operatively for Reimer's migration index (MI), acetabular depth ratio (ADR), and acetabular index (AI) to track the progression of hip instability following surgery. A piecewise linear mixed effects model was used to interpret the data, accounting for correlation within individual measurements. Demographics and relevant clinical information such as the occurrence of re-subluxation and need for revision surgery were also recorded.

RESULTS

24 patients (average age at surgery: 7.3 ± 3.4 years) were included in this analysis. 3 patients (12.5%) had residual hip subluxation (MI>30%) immediately following surgery, 11 patients (45.8%) experienced re-subluxation (MI>30%), and 4 patients (16.7%) required subsequent revision hip surgery. By 4 years following index surgery, the MI of the cohort on average measured 30% on the operative side (see Fig. 1). AI and ADR progressed in a similar fashion following surgical correction.

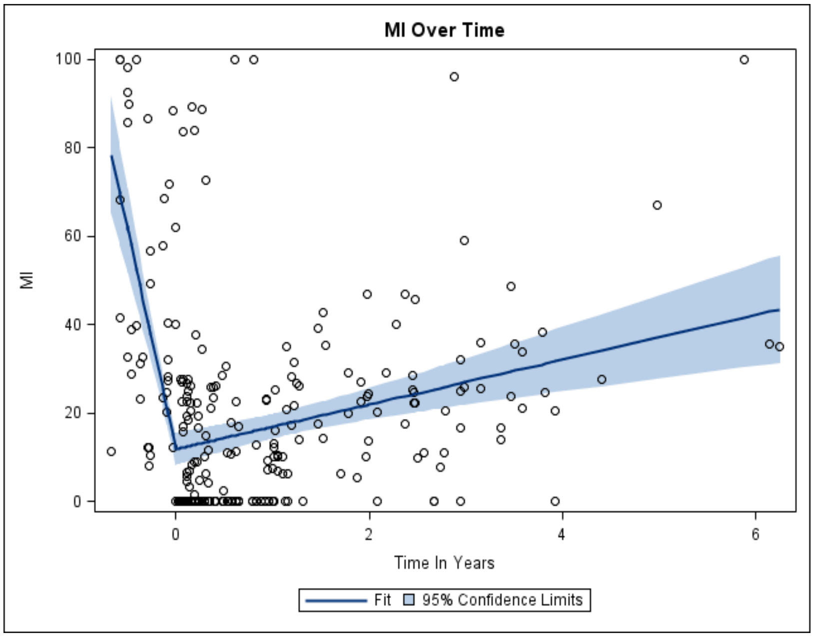

Figure 1. Migration Index (MI) over time following surgery.

DISCUSSION: Patients with syndromic disorders or chromosomal abnormalities experience a high rate of hip re-subluxation following VDRO and/or PO and are often at risk of requiring revision surgery. The etiologies of the complications are incompletely understood, and the principles of CP hip reconstruction may not apply to this cohort of patients.

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DISCLOSURE STATEMENT

No authors have any relevant financial conflicts to disclose.

A comparative assessment on utility of clinical range of motion versus radiographic imaging in predicting gait range of motion in femoroacetabular impingement

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INTRODUCTION

Contemporary gait analyses on femoroacetabular impingement syndrome (FAIS) have traditionally focused the condition can be defined with systematic variables [1]. However, little data exists on the association between disease progression and kinematic changes. We describe a novel study in which techniques and parameters for determining severity of disease, specifically radiographic imaging and clinical range of motion, are examined for their association with poor gait motion. We hypothesize that changes in clinical passive range of motion tests will be associated with greater deviations in gait motion than radiographic data in patients with FAIS.

CLINICAL SIGNIFICANCE

We expect that the study’s findings can be clinically employed to better understand burden of disease and establish methods of care tailored to individual presentations of FAIS. The findings from this study can also be used by future studies in determining what factors definitely lead to poor gait motion in FAIS.

METHODS

Twelve FAIS patients selected to undergo hip preservation surgery were analyzed for numerous radiographic parameters via radiographic, CT, and MRI imaging. These variables included lateral center edge angle, alpha angle, femoral torsion, acetabular version, spinopelvic tilt, ischiofemoral space, and head-neck offset ratio. An orthopedic surgeon then assessed the patients’ passive range of motion in flexion, internal rotation in flexion, external rotation in flexion, extension, internal rotation in extension and external rotation in extension. Finally, for gait motion, skin markers followed by an optical tracking system were used to measure sagittal peak flexion, sagittal peak extension, frontal range of motion, and transverse range of motion for symptomatic and asymptomatic hips at three different levels of incline (0°, 10°, - 10°). These variables were compared using a spearman rho test to see the association between the numerous independent variables and gait.

RESULTS

A total of 38 variables were measured in addition to basic demographics, age, and sex of each patient to correlate with gait. The average age of patients assessed was 31 and the average weight was 87 kilograms. Average values of variables that are commonly utilized for clinical judgement are presented in Table 1.

Table 1: Average values for clinically important numerical variables

Independent Parameter	AP Radiograph α-angle	CT Femoral Version	Passive Internal Rotation in Flexion
Measured Average ± S.D.	60.53° ± 16.60°	9.71° ± 5.31°	5.36° ± 8.12°
Dependent Parameter	Level Sagittal Peak Flexion	10° Incline Sagittal Peak Flexion	10° Decline Sagittal Peak Flexion
Measured Average ± S.D.	23.96° ± 4.26°	43.79° ± 4.70°	17.33° ± 4.919°

Preliminary results of the spearman rho test generally demonstrated that changes in clinical examination variables were more correlated with a greater change in gait patterns. For example, the greatest monotonic association between two variables was found between the passive flexion range of motion and sagittal peak extension at 0° incline in the symptomatic hip at a correlation coefficient of -0.88694 ($p = .01185$). Additionally, passive flexion range of motion proved to be the variable with the most statistically significant relationships with gait variables as a whole.

Results also indicated that clinically important diagnostic measurements, listed in Table 1 under independent parameters, have little to no association with gait patterns. The only significant relationship that could be drawn between the three listed variables and gait was CT femoral version and sagittal peak flexion at 10° incline in the asymptomatic hip at a correlation coefficient of -.7 ($p = .036$). However, more statistical analysis needs to be completed before full interpretations can be drawn from these results.

DISCUSSION

Results indicated that generally, use of clinical tests, such as the passive flexion range of motion test, should be considered more extensively in determining how pathology of FAIS impacts a patient's gait motion. Even though clinical data had a generally had a greater correlation with gait outcomes, radiographic data remains an essential tool to supplement diagnosis and prognosis.

Gait is one of the quintessential activities of daily living (ADLs) so it is imperative for clinicians to understand the association between gait variables and FAIS morphology if they aim to fully understand the complications of the condition. This study may facilitate greater care for FAIS patients by giving clinicians a better understanding of how commonly used methods for diagnosis and prognosis of FAIS correlate with complications.

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DISCLOSURE STATEMENT

Authors have no conflicts of interest to disclose.

GAIT ASYMMETRY IS ASSOCIATED WITH MOBILITY PERFORMANCE OF INDIVIDUALS WITH LOWER LIMB AMPUTATION

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INTRODUCTION

Lower limb amputation results in reduced mobility performance [1], which may lead to increased risk for disability, poor quality-of-life, and mortality [2]. Identifying potentially modifiable factors related to mobility performance, such as gait characteristics underpinning walking, is crucial to the development of effective rehabilitation interventions. Gait asymmetry is common after lower limb amputation, and may impact mobility performance. The relationship between gait asymmetry and mobility performance, however, is not well understood in those with lower limb amputation. Gait asymmetry in people with lower limb amputation, characterized by a shorter step length and longer time spent in stance on the intact side, has been associated with poor health outcomes, such as increased risk of intact side knee/hip osteoarthritis and falling [3]. Further, gait asymmetry may be energetically costly, as observed in the stroke population [4], and energetically inefficient walking is predictive of a steeper mobility decline over time [5]. If greater gait asymmetry is associated with poor performance on mobility-based outcomes, reducing gait asymmetry may become a critical objective during post-amputation rehabilitation. In this cross-sectional study of adults with transtibial (TTA) or transfemoral amputation (TFA), we explored relationships between gait asymmetry and mobility performance, as assessed with several clinical outcome measures.

CLINICAL SIGNIFICANCE

The findings of this study suggest post-amputation rehabilitation for people with TTA or TFA may consider targeting gait asymmetry in order to improve mobility performance.

METHODS

Included participants were aged 18-85yrs, ≥1-year post unilateral TTA or TFA, used a prosthesis to walk inside and outside the home and did not participate in regular physical activity, i.e., Level I or II on the Saltin-Grimby Physical Activity Level Scale [6].

Participants completed 3 walking trials at a fast walking speed over the GaitRite® (Gold, CIR Systems, PA, USA) system. Step length (m) and stance time (secs) of the prosthetic (P) and intact side (I) were extracted from GaitRite® software. Symmetry indices (SI), where higher values indicate greater asymmetry, were calculated for both step length and stance time using the following equation:

$$SI = \left| \frac{(I - P)}{(I + P) * 0.5} \right|$$

Mobility performance was quantified using the Timed Up and Go test (TUG) and Six-Minute Walk (6MWT).

Statistics

Demographic characteristics were examined using descriptive statistics. A hierarchical multiple linear regression model was used for each outcome (i.e., TUG and 6MWT), where time since amputation was entered in Block I, level of amputation (TTA=0; TFA=1) in Block II, and step length SI & stance time SI in Block III ($\alpha \leq .05$).

RESULTS

Thirty-eight adults (TTA=9F / 15M; TFA=5F / 9M) participated in this study, with a median age of 62.0 yrs (TTA=64.5 yrs; TFA=60.0 yrs) and median time since amputation of 8 yrs (TTA=6.0 yrs; TFA=13.0yrs).

Step length and stance time SI were significantly associated with each outcome, such that a 0.1 increase in step length or stance time SI may result in a 0.2 or 0.5 sec increase in time to perform the TUG, and 4.0 m or 7.7 m decrease in distance covered during the 6MWT, respectively (Table 2).

Table 2. Regression Models

Block	**R^2**	**R^2 Change**	***P***
Dependent variable: TUG time (sec)			
I	0.01	0.01	0.61
II	0.14	0.13	0.03
III	0.68	0.54	0.00
B: Step length SI = 0.2*; Stance time SI = 0.5*;			
Dependent variable: 6MWT distance (m)			
I	0.00	0.00	0.96
II	0.02	0.02	0.37
III	0.44	0.42	0.00
B: Step length SI = - 4.0*; Stance time SI = - 7.7*;			

Block I: Time since amputation.
Block II: Block I + Level of amputation.
Block III: Block II + Step length SI + Stance time SI.
*B: Unstandardized beta coefficient; *p < .05*

DISCUSSION

Findings suggest gait asymmetry during fast walking is associated with worse mobility performance in individuals with lower limb amputation. Thus, a clinical focus on gait asymmetry may be crucial to improving mobility performance. The study's cross-sectional nature, however, does not allow conclusions to be drawn regarding causality and results are only generalizable to older adults with unilateral lower limb amputation with lower self-reported physical activity levels.

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DISCLOSURE STATEMENT

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COMPARISON OF NEW PROSTHETIC FOOT DESIGN IN EXPERIENCED UNILATERAL TRANS-TIBIAL AMPUTEES

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INTRODUCTION

The prosthetic foot is a critical component – arguable the single most important physical component – required for optimal function and safety of the walking performance for a person with an amputation. There are a myriad of foot component choices but little objective (or even subjective) data available to help clinicians make choices about which feet to prescribe and how best to use them.

CLINICAL SIGNIFICANCE

More subjective and objective data can allow clinicians to make more informed decisions that improve outcomes as well as avoid over prescribing and thus can help to regulate unnecessary cost to the system.

METHODS

To compare Pro-Flex XC TS and Pro-Flex LP TS feet with the patient's usual foot design in a prospective study with 10 trans-tibial users. The Pro-Flex line features energy return and smooth roll-over, shock absorption and rotational capabilities with full-length toe, and a wider foot. Both subjective and/or objective data on comfort, noise, balance and walking parameters were collected during each of two visits, four weeks apart. All subjects were evaluated in the gait lab in both their original foot as well as one of the Pro-Flex feet listed above. Subject demographics were as follows: 7 male, 3 female; all were experienced prosthetic users with time since amputation ranging from 2 to 61 years; 4 had a PTB socket, 3 a total surface bearing socket and 3 other; 5 had amputations as a result of trauma, 4 due to disease; 2 users had a Pro-Flex foot, 2 had the XC Rotate and the remaining 6 had other varying feet.

RESULTS

The subjective evaluation of walking outcomes, based on the PEQ questions 13 & 14, are shown below in Figure 1. One objective measure of walking performance was the 6 Minute Walk Test. Change scores in this measure are shown in Figure 2. L-Test scores were mixed – with six subjects showing small improvements in the experimental foot and the other four showing slight decreases. Overall walking velocity showed increases in seven of the 10 subjects with an average increase of under 10% for those seven; of the three who showed a decrease, two subjects showed a greater than 20% reduction. Five of the subjects showed an increase in base of support width, two showed no change and three showed a decrease. Activities Specific Balance Confidence Scale scores showed mixed changes with five subjects favoring the Pro-Flex foot and four favoring their original feet. Question 2F of the PEQ asks, "rate how often you felt off balance while using your prosthesis". Seven subjects indicated a score that favored the Pro-Flex foot and three subjects scored their original foot more favorably.

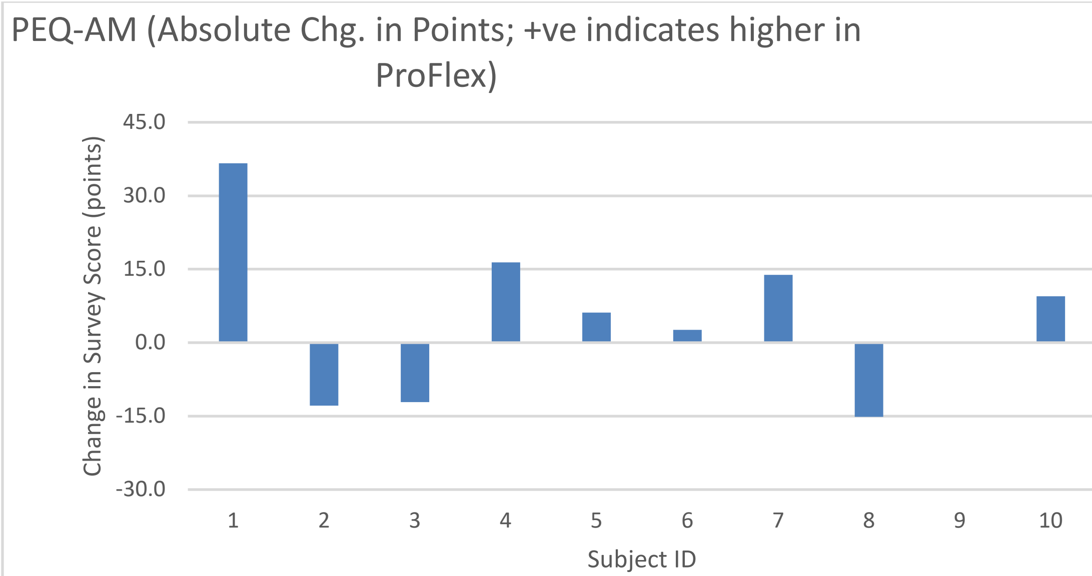

Figure 1. Subjective Scores on PEQ Questions 13 & 14 related to walking function. Higher (>0) scores favor the Pro-Flex foot.

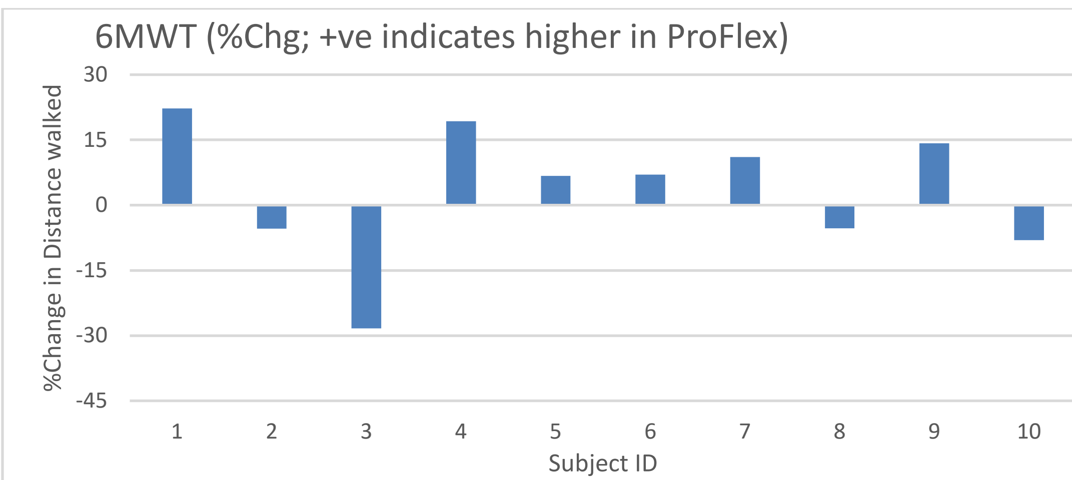

Figure 2. Change scores in the 6 Minute Walk Test. A Positive change favors the Pro-Flex Foot.

DISCUSSION

Subjects seemed to do well overall in the Pro-Flex experimental feet. PEQ questions on walking function as well as on stability were fairly strongly supportive. Though there is often some bias towards new devices and componentry, the objective data also supported the Pro-Flex foot for some of the subjects – take for example the 6Minute Walk test shown above. While statistical comparisons were difficult for such a small sample, some of the trends were interesting and worthwhile of further inquiry. Overall, this new design should be considered as an excellent choice for patients with Medicare functional ambulatory category 2 through 4. Tests of function and satisfaction should be used in future prosthetic foot trials and included in foot prescription.

ACKNOWLEDGMENTS – Barbara Hirai, Angelo Russelo and Maria Flach were invaluable to obtaining the data. Thanks to all subjects for participating.

Does prosthetic liner material affect amputee gait mechanics?

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Introduction - Use of limb prostheses for ambulation is often associated with gait deviations that make up a large part of the efficiency penalty when compared with able-bodied gait [1]. Among the parameters that have traditionally been investigated in order to address those gait deviations are prosthetic componentry, static alignment, and socket fit [2]. The material selection is an important aspect affecting socket comfort and possibly the quality of prosthesis utilization (e.g., deliberate weight bearing during the step cycle), and, by extension, gait patterns. However, these hypothesized connections have not been thoroughly investigated yet. We report preliminary findings from a randomized control trial, comparing gait variables after long-term use of regular and phase-change prosthesis liner materials [3].

Methods – Participants who use liner suspension in their lower limb prosthesis were recruited for this study. They received a pair of regular liners for use during a six-month period and a pair of phase change material liners for another six-month period in randomized sequence. Liners were produced to look identical in order to achieve blinding of participants, interventionalists, and assessors. For the purpose of comparing gait symmetry between interventions, pressure-sensitive shoe inserts (TekScan F-Scan) were used to collect data at 6-weekly assessment points. Bilateral asymmetry was calculated based on pressure profiles and plotted for comparison.

Results – Partial data from a sub- cohort (Table 1) showed differences in asymmetry (Figure 1).

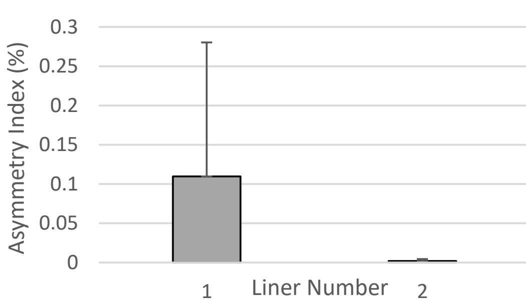

Figure 1: Asymmetry index for average stride time (0=best)

Table 1: Participant Characteristics

Participant Number	Age (Yr)	Gender (M/F)	Height (in)	Weight (lb)	Plus-M Score
1	66	M	67	190	58.4
2	42	M	77	300	47.7
5	63	M	70	176	64.5
7	36	F	60	140	67.1
8	65	M	74	255	62.5
10	58	F	70	128	61
11	61	M	67	268	67.1
12	58	M	71	150	55.3
13	68	M	69	247	52
14	67	M	70	220	55.3

Discussion – While the preliminary data do not conclusively answer the question on the effect of liner materials on gait, the found trends provide some support for the assumption that temperature control liners lead to greater socket comfort and thus increased prosthesis utilization in the sense of better gait symmetry. The study is currently ongoing, with additional data analysis expected to address the current sample size limitation.

Key words: Activity levels, Gait, Phase-change material (PCM), Prosthetic liner

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A SURVEY OF DATA COLLECTION PROTOCOLS FROM 13 MOTION ANALYSIS LABORATORIES

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INTRODUCTION

The Motion Analysis Centers in the Shriners Hospitals for Children system use three dimensional motion analysis to quantify gait metrics used to describe gait pathology, assist with treatment planning, and evaluate outcomes for children with movement disorders. A challenge when performing multicenter research is variability in data collection methods. Shriners Motion Analysis Center Network (SMACnet) recognizes this challenge and has worked to standardize data collection strategies for a clinical gait analysis by developing a minimum standard gait analysis protocol (MSGAP)[1]. After implementing this protocol among SMACnet laboratories, a study showed "an average 20% decrease in the standard deviation of 7 of 9 kinematic measures and an average 29% decrease in the maximum difference between examiners of 8 of 9 kinematic measures[2]." Human error will always be a variable, but can be minimized with the implementation of the MSGAP. However, other sources of variability that have not been as widely discussed in the literature include different hardware, software, biomechanical modelling techniques, lab environments, and patient instructions. The objectives of this report were to survey the thirteen SMACnet laboratories about their standard clinical gait analysis protocol, and compare the results.

CLINICAL SIGNFICANCE

Describing data collection protocol methods is an important step in the process of minimizing lab variability in order to develop a pediatric gait data registry using data from Shriners Motion Analysis Center Network (SMACnet).

METHODS

Investigators for this project surveyed 13 motion laboratories in SMACnet by emailing a spreadsheet containing questions about data collection protocols.

RESULTS

Table 1. Clinical Gait Analysis Protocol - Survey Results	**# of Labs out of 13**		**# of Labs out of 13**
Motion Capture System – Vicon	12	Anthropometric Measures:	
Motion Capture System - Motion Analysis Corporation	1	Height & Weight	13
≥10 Motion Capture Cameras	13	Leg Length	13
Motion Capture Camera Frame Rate 240 Hz	2	Knee Width	11
Motion Capture Camera Frame Rate 120 Hz	4	Ankle Width	11
Motion Capture Camera Frame Rate 100 Hz	3	ASIS to ASIS	8
Motion Capture Camera Frame Rate 60 Hz	1	Tibial Torsion	8
Collects Reference Video	13	Ankle Varus/Valgus Angle	7
Marker Trajectory Filter – Woltring	8	ASIS to Greater Trochanter	4
Marker Trajectory Filter – Butterworth	4	Pelvic/trunk Rotation Sequence - TOR	8
Marker Trajectory Filter – 3rd Order Polynomial	1	Pelvic/trunk Rotation Sequence - ROT	5
Reflective Marker Size 14mm	8	Identify Gait Events Manually	5
Reflective Marker Size 9.5mm	4	Identify Gait Events Automatically from FP Strikes	8
Reflective Marker Size 6.4mm	1	Use Autocorrelate to help identify events	2
≥2 Force Plates	13	Lab Software – Vicon Nexus	12
Foot Model – Bruening	1	Lab Software – C-Motion Visual 3D	3
Foot Model – Milwaukee	1	Lab Software – Motion Analysis Patients & Services (MAPS)	8
Foot Model – mSHCG	7	Custom Software	5

The results of this survey can be seen in Tables 1 and 2. Additionally, laboratory staff were asked in what ways they document each gait trial collected. All labs reported recording if the trial is 'Static' or 'Dynamic', if the feet are plantigrade, if the subject is barefoot, shod, braced and the assistive devices used. In regards to hip joint modelling 2 labs reported using the Harrington Hip model, 4 labs measure and input the ASIS to greater trochanter distance and 7 labs use Nexus' built in regression equations to calculate hip joint center. Three out of 13 labs use a virtual pointer (e.g. Dypstick) and software for creating virtual markers on the pelvis for patients where soft tissue artifact can impact data validity. The data collection sequence varied among labs for the order in which observational video, physical therapist measurements, pedobabarography, VO2 testing, running, braced gait and barefoot gait analysis were performed.

Table 2. Marker Placement	**# of Labs out of 13**	**Marker Placement**	**# of Labs out of 13**
Head	4	Patella	11
C7	11	Thigh Triad	2
Sternoclavicular	13	Knee Alignment Device (Static)	10
T10	2	Lateral Knee	13
Shoulder	6	Tibial Lateral Wand	6
Elbow	6	Tibial Lateral Triad	2
Wrist	3	Tibial Anterior Crest	5
Anterior Superior Illiac Spine	13	Lateral Ankle	13
Sacrum	9	Heel	13
Posterior Superior Illiac Spine	9	Toe 2/3rd Metatarsal	13

DISCUSSION

This report is part of an effort to investigate data compatibility among SMACnet laboratories in order to identify different data collection methods and test whether those differences create compatible data or if certain levels of standardization are necessary. We used a survey to gather data about lab protocols from 13 labs in SMACnet. With this data we can better understand the sources of variability and task experts with deciding on what protocols maximize compatibility without stifling innovation. Note that this data represents typical protocols and potential variations exists depending on the patient. Based on this report we feel important topic areas for discussion include: pelvic/trunk rotation sequences, hip modelling, data capture frequency, marker locations on the tibia and data filtering. We hope that this report will help stimulate further discussion on best practices and establish lab protocols that lead to the highest quality and most compatible data possible.

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ACKNOWLEDGMENTS

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DISCLOSURE STATEMENT

Nothing to disclose.

A COMPARISON OF KINEMATIC-BASED FOOT VELOCITY, SHANK ANGULAR VELOCITY, COORDINATE-BASED TREADMILL ALGORITHMS VERSES KINETIC FORCE PLATE DATA IN DETECTING HEEL-STRIKE AND TOE-OFF CHILDREN WITH AND WITHOUT CEREBRAL PALSY, AND UNIMPAIRED ADULTS

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INTRODUCTION

There is a boom in the development of video and wearable sensor-based techniques for gait analysis outside of the typical instrumented motion analysis laboratory. Such systems do not use kinetic data from force plates in their algorithms to determine gait events such as heel-strike and toe-off. Such systems make estimations of these gait events from video data or sensor signals, such as those from gyroscopes or accelerometers. Ultimately, inflections in a derived or measured variable, e.g., the velocity of a limb segment, are used to define the specific gait events.

CLINICAL SIGNIFICANCE

It is important to determine if video and sensor derived gait event data are valid, reliable and robust when compared to laboratory based measures that use kinetic data from force plates. This study aims at evaluating and comparing the accuracy of three widely used kinematic methods, Foot Velocity Algorithm (FVA), Shank Angular Velocity (SK), Coordinate Based Treadmill Algorithm (CBTA) versus the kinetic method using an instrumented treadmill's force plates to detect heel-strike and toe-off in children with cerebral palsy (CP).

METHODS

Walking gait analysis of 6 children with CP, age 12-18y, GMFCS I – III; 5 typical developing kids, age 10-16y, and 6 adults was performed at University of Delaware and Shriners Hospital for Children, Philadelphia. For this observational research design, all participants signed consent forms approved by the institution's Human Subjects Review Board. An eight-camera system (Motion Analysis Corporation, Santa Rosa, CA), with sampling frequency of 128 Hz, and two force platforms (Bertec Corporation, Columbus, OH), with a sampling frequency of 3200 Hz, were used to record 30 seconds of walking data. Data were processed using the three different kinematic methods (FVA, SK, CBTA) in Visual 3D and compared to kinetic force plate data.

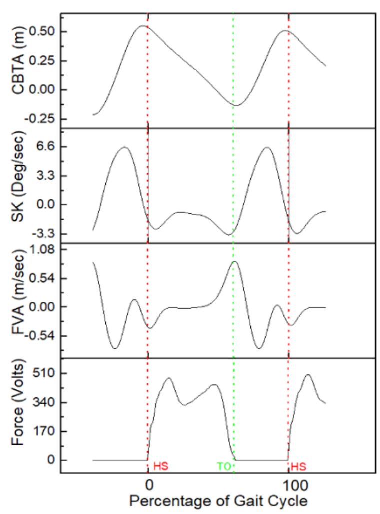

Figure 1: The three kinematic and kinetic methods indicating toe-off (TO) and heel-strike (HS) while an individual walk.

FVA uses the vertical velocity of the foot center to detect heel-strike and toe-off with troughs and peaks representing heel-strike and toe-off, respectively [1]. The SK method uses the negative zero crossing of shank angular velocity to detect heel-strike and the second minimum value to detect toe-off [2]. CBTA plots the sinusoidal curve formed by the X coordinate of the foot maker as a function of time of which, the peaks and valleys represent heel-strike and toe-off [3] (Fig. 1). For kinetic methods using force plates, HS is represented by the initiation of force detection upon contact, and toe-off as the return of force to zero as contact with the plate ceases.Heel-strike and toe-off identified using the three kinematic algorithms were fit to those detected by the force plate data using a linear mixed (random coefficients) model. Root mean score errors (RMSEs) for each method were calculated. Heel-strike and toe-off detection rates were defined as the ratio of the number of events detected by the kinematic algorithms to those detected by the force plate data, and reported as a percentage for each algorithm.

RESULTS

The RMSE values for all subjects for each method were 0.1164, 0.0432, 0.0315 for FVA, SK and CBTA, respectively (p<0.0001). A group effect was seen only in CBTA, where CP had significantly higher residuals than typical developing kids and adults (p=0.0051). The heel-strike detection rate in CP was 88%, 94% and 100% and the toe-off detection rate was 95%, 95%, 95%, for FVA, SK, and CBTA, respectively. The distribution of residuals for all subjects from the three methods depicted one individual with the largest residuals (*Figure 2: blue dots*). Further investigation showed that this subject was an individual with CP having a particularly impaired crouch gait, a clear outlier in the RMSE data. When omitting this individual's data, the RMSE values for each method were 0.06997, 0.0.0241, 0.0200 for FVA, SK and CBTA, respectively (p<0.0001). CP had higher residuals (p=0.0452).

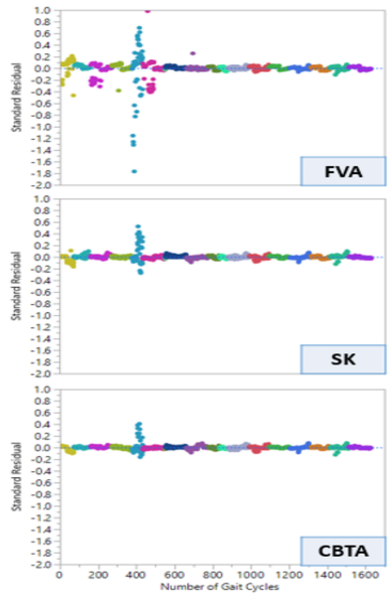

Figure 2: The distribution of residuals for each individual across FVA, SK and CBTA.

DISCUSSION

Linear mixed random coefficients models showed that all the methods did an excellent job in detecting the gait events in our subjects. Overall, CBTA was the best with lower RMSEs and higher detection rates, followed by SK and FVA. The CBTA requires a fully equipped motion capture gait lab, however, SK can be used portably outside of the laboratory setting in which shank angular velocity is derived from video or wearable sensor data. Due to the larger residuals and lower detection rate of heel-strike, FVA might not be the best method to be used for determining gait events in children with CP. Further, individuals with severe crouch gait impairments may require the use of fully instrumented labs rather than sensors for accurate and reliable gait phase detection.

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DISCLOSURE STATEMENT

There are no conflicts of interest to disclose.

PERFORMANCE ANALYSIS OF GOLF SWING IN DIFFERENT DISTANCES AND CLUBS USING WIRELESS INERTIAL MEASUREMENT UNIT SENSOR

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INTRODUCTION

Swing tempo and rhythm are mainly described by the golfer's swing skill and shown to be important in maintaining the performance. The tempo means the time required for the swing from the address to the impact. Rhythm means the ratio of the downswing to the backswing.

The wearable IMU sensors have been widely utilized in the motion analysis during sports [1], and also recently getting more popularized in the study of golf swing analysis [2,3]. In this study, we analyzed the golf swing performance, swing tempo and rhythm using wireless IMU sensor. First, we validate the estimated clubhead trajectory from the IMU sensor with data from the optical motion capture system during the golf swing. Then, the swing performance parameters were represented to automatically detecting from the motion data. After that, different golf club swing including a driver, 7-iron, and wedge with 30 m, 50 m, 70 m, and full swing were applied to calculate the golf swing tempo and rhythms.

METHODS

In order to estimate the clubhead trajectory using 3-axis acceleration data and the 3-axis gyroscope data from the IMU sensor (Wearnotch®, Notch Interfaces Inc., New Jersey, USA), an in-house modified Madwick filtering algorithm was developed using Matlab® R2015a (The Mathworks Inc., USA). The estimated clubhead trajectory was validated against data from the optical motion capture system with 10 cameras (Hawk® system, Motion Analysis, CA, USA).

The IMU sensor was fixed to the near to the club-grip, and the three optical markers were attached to the middle of the golf club (Fig 1a). In the experiment, 14 college-level male golfers (age, 21.5 ± 2.1 years; handicap, 6.7 ± 4.8) were recruited from the Korean College Golf Federation. Each subject repeatedly performed the six different golf swings; driver full swing, 7-iron full swing, wedge full swing, wedge 30m swing, wedge 50m swing, and wedge 70m swing, respectively. The swing tempo is calculated with the time from address to ball impact. The swing rhythm is calculated as the ratio of the downswing to backswing. Each parameter were calculated based on the estimated clubhead trajectories during the golf swing.

RESULTS

The estimated clubhead trajectory using the IMU sensor showed good agreement with the data from the optical motion camera system (Fig 1b, and 1c). The Pearson correlation coefficients and RMSE values were 0.89-0.95 and 0.11-0.17.

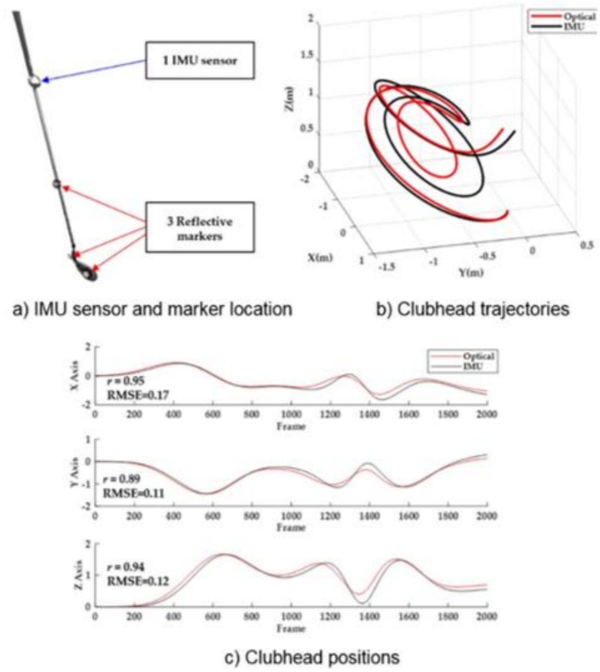

Figure 1: Average of clubhead trajectory and position data from the IMU sensor and optical motion camera.

The swing tempo was 1.14 ± 0.14 (s), 1.13 ± 0.16 (s), and 1.13 ± 0.15 (s) for the driver, 7-iron, and wedge, respectively. The swing tempos of all the participants were presented very consistently in a range from 1.13 to 1.14, irrespective of club type ($p = 0.90$). The swing tempo was 1.08 ± 0.14 (s), 1.06 ± 0.12 (s), and 0.95 ± 0.07 (s) for the 70-m wedge, 50-m wedge, and 30-m wedge, respectively. The swing tempos at 70-m and 50-m were similar but the tempo at 30-m was shorter ($p < 0.05$). Two-way ANOVA test showed a significant difference between the clubs and distances groups ($p < 0.05$).

The swing rhythm was 4.25± 0.73, 4.00 ± 0.80, and 3.95 ± 0.83 for the driver, 7-iron, and wedge, respectively. The swing rhythm tends to decrease for club types with less carrying distance ($p = 0.51$). The swing rhythm was 3.43 ± 0.79, 2.95 ± 0.63, and 2.35 ± 0.33 for the 70-m wedge, 50-m wedge, and 30-m wedge, respectively ($p < 0.05$). Similarly, the significant difference was observed between the two groups ($p < 0.05$).

DISCUSSION

The wearable IMU sensor has distinct advantages such as no limitations on the experimental space, shorter data receiving and processing time, and convenient wearability (to expand to other types of wearable devices). In our experiment, the total data receiving and processing time was several seconds, meaning that the present technology can be applied to practical fields. All the golf performance parameters were successfully calculated and presented to the players nearly instantly, so that the players or coaches could use them.

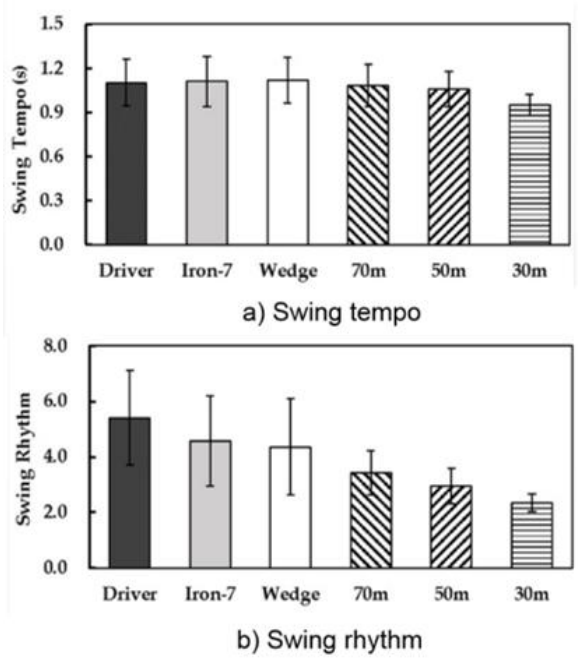

Figure 2: The swing tempo (a) and rhythm (b) for different golf club and distances.

The measured swing tempo were 1.1 s to 1.2 s for full swings with the driver, 7-iron, and wedge. Statistical analysis showed that there were no differences between different clubs used for full swings. In contrast, the tempo was significantly different between swings for short distances. Moreover, there was a significant difference between the clubs and distances groups on the tempo. These results were similar to previous studies.

The swing rhythm is used as a swing speed indicator which refers to the overall speed of the entire sequence. Generally, the backswing phase is a longer period of time and the downswing phase is a shorter period of time. In our study, the swing rhythm tended to slightly decrease from driver to 7-iron to full-wedge swing, which is in accordance with the decrease of the carrying distance. In addition, the swing rhythm decreased as the carrying distance of the wedge swing decreased from 70 m to 30 m.

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DISCLOSURE STATEMENT

All authors have nothing to disclose.

Short and Long Term Effect of Scoliosis Bracing on Pain and Function in Adult Degenerative Scoliosis Patients

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INTRODUCTION

Adult degenerative scoliosis (ADS) results from age-related changes leading to segmental instability, deformity and stenosis. Although the etiology is unclear, degenerative adult scoliosis is associated with progressive and asymmetric degeneration of the disc and facet joints, which typically leads to stenosis.[1] By virtue of the narrowed spinal canal associated with the degeneration these patients frequently develop back pain, as well as leg pain, weakness, and numbness.[1] With an aging population in the USA and increased attention to the quality of life versus cost issues in the current healthcare environment, degenerative adult scoliosis has become a considerable healthcare concern.[1] Patients with scoliosis demonstrate an altered gait pattern.[2] Such differences include decreased step length and reduced range of motion in the upper and lower extremities[2], asymmetry of trunk rotation and ground reaction force in three-dimensions. Mahaudens et al.[3] found a decrease in the muscular mechanical work associated with an increase in energy cost and a decrease in the muscular efficiency in a scoliosis population compared to healthy controls. Furthermore, scoliosis patients exert 30% more physical effort than healthy subjects to ensure habitual locomotion, and this additional effort requires a reciprocal increase in oxygen consumption. This altered gait pattern demonstrated by subjects with scoliosis may be due to changes in global postural control strategies caused by pain and spinal deformity.[2] The timed-up-and-go (TUG) and 6min walking distance tests have previously been shown to be strongly tied to improved functional abilities and quality of daily life. Previous research showed that scoliosis patients do not have impaired postural balance when compared to healthy controls[4], while several others did find an effect of scoliosis on postural balance.[5] This discrepancy in findings may be due to differences in curve characteristics included and their effects on postural balance, curve types (single or double), the number of different curve types, location of curves (thoracic and lumbar), and/or Cobb angles.[5] A lumbar-sacral orthosis (LSO) has been found to reduce pain within a short time in ADS patients.[6] Custom-made rigid torso braces, similar to those commonly used for children, are sometimes used in ADS patients; however, only anecdotal evidence of their efficacy is available[6] and problems with comfort and compliance are quite frequent. A new brace has recently become available, scoliosis supporting LSO, designed to alleviate pain in adult patients with chronic pain secondary to scoliosis. The scoliosis LSO led to some improvement of pain at 1 month in a group of adult women with scoliosis and chronic low back pain, but the quality of life did not change significantly.[6] This is may be due to the very short follow-up time. The effect of these braces on functional tasks and activities of daily living including walking have not been studied either. The purpose of this study is to determine if the use of a scoliosis-supporting LSO for ADS can produce short-term and long-term improvements in gait performance and symmetry.

CLINICAL SIGNIFICANCE

The scoliosis-supporting LSO may provide additional trunk support which in turn reduces pain and allows for more natural and symmetric gait.

METHODS

This study will be a repeated measurement design. This study will be a non-randomized, prospective, concurrent control cohort study of patients with ADS who are clinically indicated for brace intervention and served as their own controls via their own pre-treatment evaluation. Twenty-five symptomatic ADS patients were included in the study (Age:

74.2±1.9; Height: 1.6±0.1m; Weight: 64.3±16.4kg; Cobb > 25°) and were evaluated at three time points: before wearing the brace (Pre), after an initial period of wearing the brace for 45min (Post45m), and eight weeks after wearing the brace for a minimum of 4hrs a day (Post8w). All test subjects were fitted with a full-body reflective marker set. During each evaluation, patients completed walking trials at a self-selected speed, balance test, a TUG test, a 6min walking test, and patients reported outcomes (PROMs) including visual analog scale (VAS), Oswestry Disability Index (ODI), and SRS22r. Three-dimensional motion tracking was used to collect spatiotemporal data. Three-dimensional (3D) kinematic data was recorded at 120Hz via a human motion capture (Vicon Nexus 2.0 Inc., Englewood, CO). Ground reaction force (GRF) (AMTI Corp, Watertown, MA.) data was recorded simultaneously at 1200 Hz. The kinematic data was low pass filtered with a 4th order Butterworth filter with a lower cutoff at 4Hz. The GRF data was also low pass filtered FFT with a similar filtering technique. Spine, pelvis, hip, knee, and ankle kinematics and GRF was analyzed using a custom MATLAB program. Outcome measures were pelvis and lower-extremity RoM, spatiotemporal gait parameters, and foot GRF, TUG, 6min walking distance, and PROMs. Repeated-measure analysis of variance (ANOVA) and post-hoc paired t-tests were used to test for differences between short-term (Pre to Post45m) and long-term (Post45m to Post8w)

RESULTS

Patients reported a significant reduction in both VAS low-back and leg pain from Pre to Post45m (p=0.011 and p=0.029 respectively) however this was not sustained between Post45m to Post8w. A significant improvement in VAS low-back pain was seen from Pre to Post8w however this did not reach significance (p=0.038). Moreover, notable improvements were seen in ODI, and SRS22r, however, these did not reach significance. A significant increase was seen in axial pelvic RoM (p=0.01) from Pre to Post8w. A significant increase in coronal pelvic RoM was also seen from Pre to Post8w (p=0.022). A notable trend in increasing walking speed was seen from Pre to Post45m and Post45m to Post8w however the effect was not significant. ADS patients exhibited significant differences in peak GRF between right and left sides at Pre (p=0.007). At both post45 and post8w, no significant differences between right and left side GRF were seen. With the use of the scoliosis brace, ADS patients were able to complete the TUG significantly faster in the short- (Pre: 12.18s vs. Post45m: 10.75s, p=0.007) and long-term (Post8w: 9.50s, p=0.014). Patients also covered significantly longer distances in the 6-minutes walking test in the short- (Pre: 303.69 vs. Post45m: 346.93m, p<0.001) and long-term (Post8w: 374.72m, p<0.001).

DISCUSSION

Symptomatic ADS patients were able to walk significantly faster and for longer distances following both short-term and long-term use of the scoliosis-supporting LSO. Key findings included a rebalancing of GRF between sides after wearing the brace both at Post45m and Post8w and in increased pelvic RoM. Moreover, this study indicates that the use of the scoliosis-supporting LSO may be providing trunk support which in turn may be reducing pain and improving patient functional level. The sustained improvements of these measures among our patients suggest that the scoliosis-supporting LSO may serve as a valuable method for nonoperative treatment of ADS pain and reduce disability. Additionally, trends in the data suggest improved function and reduced disability. These findings suggest that the scoliosis-supporting LSO may provide additional trunk support which in turn reduces pain and allows for more natural and symmetric gait.

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DISCLOSURE STATEMENT

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Spine Motion Evaluation of a Pediatric Patient with Adolescent Idiopathic Scoliosis Undergoing Posterior Spinal Fusion: A Case Report

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PATIENT HISTORY

Patient is an 11-year old female who was diagnosed with adolescent idiopathic scoliosis approximately 3 years ago. Patients mother was the first to notice the curvature of her back. They presented to their pediatrician who then referred them to an orthopedic surgeon. Patient has a family history of scoliosis but does not have any prior medical history or history of hospitalizations.

CLINICAL DATA

Patient exam shows significant right thoracic prominence with significant rotation and forward flexion. There is also a mild right torso shift as well as mild accentuated thoracic kyphosis.

Motion DATA

Motion capture data was collected where the patient was instrumented with trunk markers to create a pelvis segment, lumbar segment, and thoracic segment. She was instructed to stand in a comfortable upright position and to actively move the trunk to the maximum end range of motion (forward flexion, backward extension right/left lateral bending, right/left axial rotation)to measure the motion between these segments. During the standing forward flexion patient seemed to present with asymmetric reduced thoracic flexion (table 1). She also showed prominent lateral right bend suggesting structural and morphological changes (table 2).

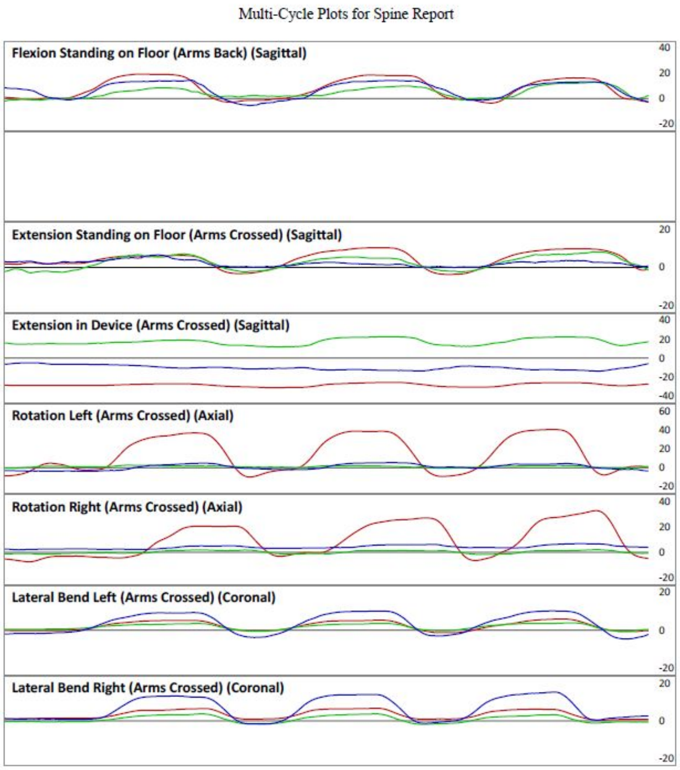

Figure1: This graph illustrates a gait analysis report of trunk range of motion.

Table 1: Flexion Standing on Floor (Arms Back) (3 cycles)

FLX FLOOR	Pre-Sx		
	Sagittal	Coronal	Axial
Thoracic (wrt lumbar)	**-6° ±4.2**	7° ±0.8	-5° ±0.4
Lumbar (wrt pelvis)	**68° ±3.2**	-5° ±0.7	-3° ±1.8
Single Seg. (wrt pelvis)	**62° ±6.6**	7° ±0.6	-10° ±1.6
Pelvis (wrt lab)	**25° ±1.4**	0° ±0.9	5° ±2.4
C7 to Floor %Δ in Vertical†:	**18%**		

† Percent is the percentage reduction in the static distance when flexed.

Table 2: Lateral Bend Right(Arms Crossed): (3 cycles)

LAT BEND R	Pre-Sx		
	Sagittal	**Coronal**	Axial
Thoracic (wrt lumbar)	38° ±1.5	**16° ±1.7**	-13° ±3.8
Lumbar (wrt pelvis)	-16° ±1.9	**18° ±0.4**	3° ±1.8
Single Seg. (wrt pelvis)	17° ±1.0	**38° ±1.2**	-7° ±2.4
Pelvis (wrt lab)	-5° ±1.1	**16° ±1.0**	18° ±2.6
R Shoulder to Floor %Δ in Vert†:	**12%**		

† Percent is the percentage reduction in the static distance when flexed.

TREATMENT DECISIONS AND INDICATIONS

In this case the patients curve is approximately 75 degrees with a compensatory left lumbar curve of about 50 degrees. Therefore, posterior spinal fusion is recommended to prevent further progression. The goal of the surgery would be for partial reduction in the curvature and for long-term prevention of continued curve progression. Surgery corrected her scoliosis.

OUTCOME

Patient is in 11 months post surgical intervention. We plan to continue following her in the Motion Analysis Center(MAC) for her 2 year postoperative evaluation. Patitent underwent a T2-L3 posterior spinal fusion. She is following the post-op spine precautions of no bending, lifting or twisting.

SUMMARY

This case study exhibits a successful treatment and management of a patient with adolescent idiopathic scoliosis. Patient has no complaints, is attending school full time and is showing promising recovery.

DISCLOSURE STATEMENT

We have no conflicts of interest to disclose.

POST-CVA UPPER LIMB 3D MOTION ANALYSIS: OCCUPATIONAL PERFORMANCE COMPARING THREE INTERVENTIONS: CASE STUDY

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PARTICIPANT HISTORY

The participant was a 55-year-old male with a history of cerebrovascular accident affecting his left upper limb.

CLINICAL DATA

The participant read and signed the informed consent form, completed an intake questionnaire regarding general health and activities of daily living.

Members of the research team checked vital signs, administered the Modified Ashworth Scale (Ashworth, 1964) and Wolf Motor Function Test (Wolf et al. 2006): See Table 1.

Wolf Motor Function Test	Post-CVA	control	Post-CVA	control
	Time (s)		Functional Ability /5	
2. Forearm to box (side)	2.49	0.42	3	5
5. Hand to table (front)	2.63	0.45	3	5
8. Reach and retrieve	5.16	0.65	3	5
9. Lift can	16.30	1.14	2	5

Table 1. Relevant Wolf Motor Function Test scores vs. age-matched healthy control

MOTION DATA

A 9-camera Qualisys Oqus™ 3D motion capture system (100 Hz) tracked head, trunk, arms, forearms, and hands to monitor shoulder, elbow, forearm, and wrist joints. 16-channel Delsys TrignoTM surface EMG muscle activity (2000 Hz) of muscles crossing the shoulders, elbows, and wrists. The participant performed 3 repetitions with the less-affected upper limb, then 3 with the more-affected side of each of the following 3 occupation tasks: beginning with the water bottle held in the hand at the lateral side of the thigh: move the bottle to a waist level shelf, move the bottle to a shoulder level shelf, bring the bottle to mouth (Figure 1). The motion analysis of these tasks was repeated following a 30-minute session of one of the following interventions: modified Constraint Induced Movement Therapy (mCIMT), Neuromuscular Electrical Stimulation (NMES) and Neurodevelopmental Therapy (NDT). This study focused on within-session effects

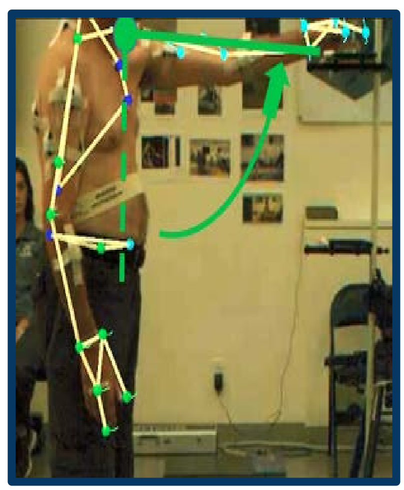

Figure 1: Participant holding bottle at shoulder height shelf: freeze frame of lateral video with three-dimensional stick figure overlay, shoulder flexion indicated in green.

Outcome Measures of note in this abstract: shoulder flexion, shoulder abduction, elbow flexion during raise to shoulder task, pre- and post-intervention (Figure 2).

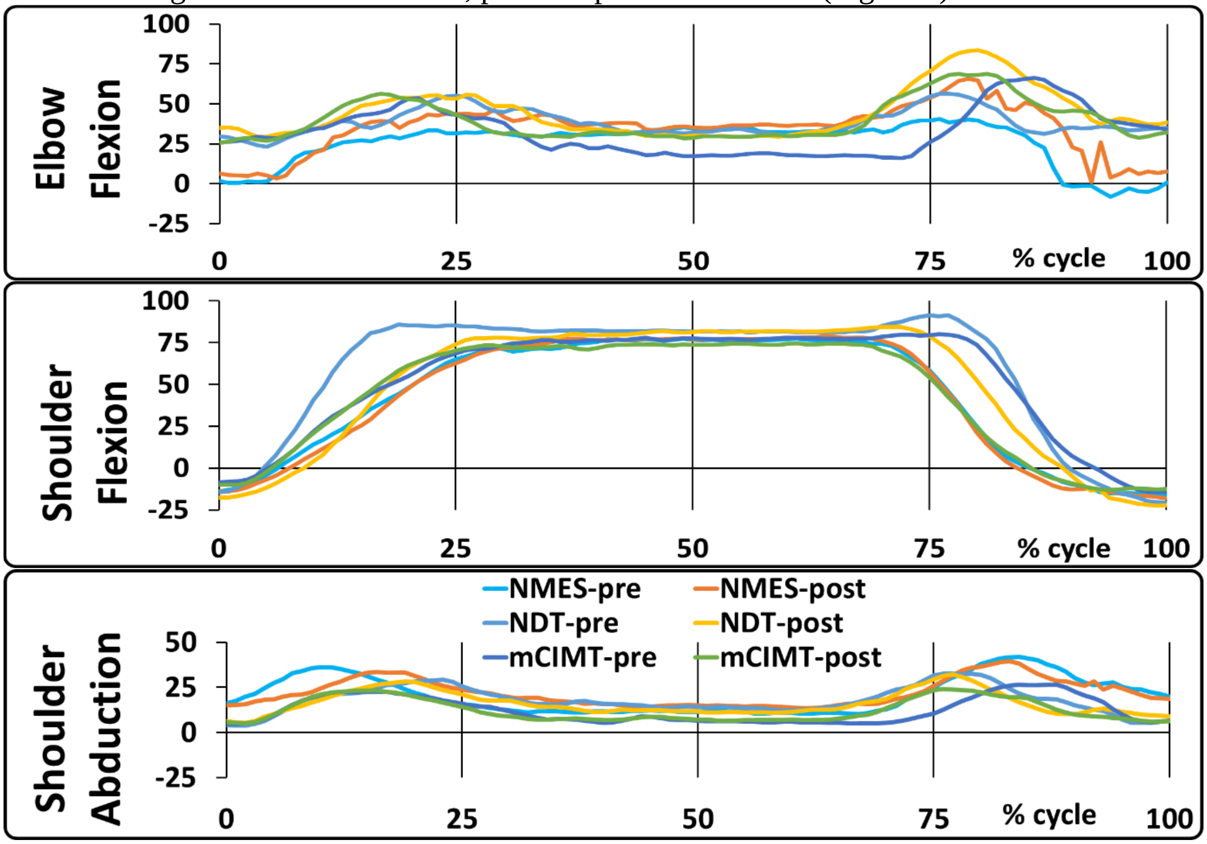

Figure 2: Average kinematics normalized to cycle: 0%: holding bottle in hand at lateral thigh, 50%: holding bottle on shoulder height shelf (see photo), 100% returning bottle in hand to lateral thigh. Comparison of pre and post NMES, NDT, and mCIMT.

TREATMENT DECISIONS AND INDICATIONS

The participant exhibited limitations in peak shoulder flexion during the reach to the shelf, compensating with increased shoulder abduction.

NMES appeared to show most improvement, closer to control participant. Post-CVA Participant preferred mCIMT.

SUMMARY

This study examined within session effects of selected interventions to address limitations in upper limb. It demonstrates the application in Occupation Therapy of 3D kinematics to quantify upper limb function, which is commonly assessed using visual observation, manual goniometers, or regular single camera video recording.

DISCLOSURE STATEMENT

The authors have no conflicts of interest to disclose.

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