Time	Wednesday, March 11th, 2009
7:00 –8:00 am	Breakfast –*Exhibitor Area*
7:00 –8:00 am	Sunrise #1 –*Room 4* "SAMSA: Standard Assessment of Motion System Accuracy" Nancy Denniston and Howard Hillstrom
8:00 –8:15 am	Welcome: James Carollo & Michael Schwartz – *Rooms 4–7*
8:15 –9:15 am	Keynote Address #1 –*Rooms 4–7* "What Do Rubber Bands, Penguins, and Kangaroos Have in Common?" Rodger Kram, PhD, University of Colorado, Boulder 1 CME/CEU Credit
9:15 –10:25 am	Podium Presentations –*Rooms 4–7* *Session 1: Orthoses and Prostheses* Moderators: Adam Rozumalski & Tishya Wren 1 CME/CEU Credit

- Effectiveness of a New Supination Control Orthotic for the Flexible Flat Foot
  W. J. Hurd, S. J. Kavros, K. R. Kaufman
- Use of the Floor Reaction Ankle Foot Orthosis in Children with Cerebral Palsy: Efficacy and Indications Based Upon Quantitative Gait Analysis
  B. M. Rogozinski, DPT; J. R. Davids, MD; R. B. Davis, PhD; G.G. Jameson, MA; D. W. Blackhurst, DrPH
- Evaluating Mechanical Ankle Foot Orthosis Characteristics Using BRUCE: A Reliability Study
  D.J. Bregman, A. Rozumalski, D. Koops, V. de Groot, M.H. Schwartz, J. Harlaar
- Short and Longer Term Changes in Amputee Walking Patterns Due to Increased Prosthesis Inertia
  J. D. Smith, P. E. Martin
- Walking on Gravel Affects Variability and Dynamic Stability Differently in Non-Impaired Subjects and Traumatic Amputees
  J. Dingwell, S.J. Scott, J. M. Wilken
- Mediolateral Ground Reaction Impulses, Stride Length and Effective Limb Length of Amputees Turning vs. Nonamputees
  A. Segal, MS; H. Sutton, BS; M. Orendurff, MS; J. Czerniecki, MD; G. Klute, PhD
- Lower Extremity Rotation Moments for Amputees Turning Compared to Nonamputees
  A. Segal, MS; H. Sutton, BS; M. Orendurff, MS; J. Czerniecki, MD; G. Klute, PhD

10:25 –11:05 am	Break: Snacks in Exhibitor Area
11:05 am –12:05 pm	Podium Presentations –*Rooms 4–7* *Session 2: Treatment Planning and Outcome* Moderators: Anita Bagley & Marjolein van der Krogt 1 CME/CEU Credit

- Gastrocnemius Muscle Tendon Unit Length: Clinical Assessment Versus Kinematic Assessment in Diabetic Subjects
  D. Morgenroth, M. Orendurff, G. Bernatz, J. Schoen, E. Rohr, J. Czerniecki
- Efficacy of Botulinum Toxine A Treatment in Children with Cerebral Palsy is Defined by Crucial Factors Within the Treatment Strategy
  K. Desloovere, PhD; A. Van Campenhout, MD; K. Fagard, PT; K. Daniels, PT; V. Schörkhuber, PT; J. De Cat, PT; D. Severijns, PT; G. Molenaers, MD, PhD
- Explicit Kinematic Criteria Predict Postoperative Improvements Following Surgical Lengthening of the Psoas Tendon in Patients with Cerebral Palsy
  A. Rozumalski, MS; W. H. Truong, MD; M. H. Schwartz, PhD; C. Beattie, PT; T. F. Novacheck, MD
- Distal Tibial Rotation Osteotomies Normalize Frontal Plane Knee Moments
  B. MacWilliams, M. McMulkin, G. Baird, P. Stevens
- Long-Term Outcomes of Orthopedic Surgery in Persons with Cerebral Palsy: Femoral Derotation Osteotomies

S. Õunpuu, MSc.; C. Nomides, DPT; M. Westwell, PT, MS; D. Roller, PT; K. Bell, MS; D. Tyburski; M. Solomito, BSBE; P. DeLuca, MD; E. Weber, MD • Cost Comparison of Initial and Subsequent Surgeries Between Patients with and without Pre-Operative Gait Analysis *T. A. L. Wren, M. Kalisvaart, C. Ghatan, S. A. Rethlefsen, R. Hara, M. Sheng, L. S. Chan, R. M. Kay*	
12:05 –1:10 pm	Commission for Motion Laboratory Accreditation Workshop *Peak Lounge, 27th Floor*
1:10 –2:30 pm	Podium Presentations –*Rooms 4–7* *Session 3: Orthopaedics* Moderators: Pam Thomason & Noelle Moreau 1 CME/CEU Credit
• The Effect of Static Standing Posture on Dynamic Walking Kinematics: Comparison of a Patella Marker vs a Thigh Wand *M. L. McMulkin, PhD; A. B. Gordon, MPT* • Gait Patterns in Children with Limb Length Discrepancy *M. Aiona, K. P. Do, K. Emara, R. Dorociak, R. Pierce* • Gait Asymmetry Following THA - Anterior vs Lateral Approach *V. Lugade, A. Wu, L.-S. Chou* • Functional Assessment of Stair Negotiation Following Total and Resurfacing Hip Arthroplasty *M. Bhowmik-Stoker, M. W. Shrader, M. C. Jacofsky* • The Role of Unopposed Quadriceps Forces in Loading the Anterior Cruciate Ligament During Jump Landing *J. Hashemi, R. Breighner, T.H. Jang, N. Chandrashekar, S. Ekwaro-Osire, J. R. Slauterbeck* • A Comparison by Motion Analysis of Anterior Cruciate Ligament Reconstruction Using a Patellar Tendon vs a Hamstring Tendon Autograft *A.L. Aguinaldo, H.G. Chambers* • Prehab Effects on Sit-to-Stand in Knee OA Patients Prior to TKA *P. M. Quesada, N. Campbell-Kyureghyan, A. M. Swank, R. V. Topp* • Effects of Gait Training with Real-Time Biofeedback on Knee Adduction Moments During Walking in Adults With Knee OA *P. Teran-Yengle, N. A. Segal, N. F. Johnston, B. Singh, J. C. Torner, R. Walace, H. J. Yack*	
2:30 –3:00 pm	Break: Snacks in Exhibitor Area
3:00 –6:00 pm	**Exhibitor Area Open**
3:00 –4:00 pm	**Poster Odd Attended –Sponsored by Vicon** ***Foyer***
4:00 –5:00 pm	**Poster Even Attended –Sponsored by Vicon** ***Foyer***
5:00 –6:00 pm	**Student Mixer – Sponsored by Innovative Sports Training** (back of Room 4)
6:00 –8:00 pm	**User Group Meetings: Motion Analysis** (Room 5-6) **Vicon** (Bubba Gump's)

EFFECTIVENESS OF A NEW SUPINATION CONTROL ORTHOTIC FOR THE FLEXIBLE FLAT FOOT

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INTRODUCTION

The hyper-pronated, flexible flat foot has been identified as a contributing factor to lower extremity injuries.[1] Orthotic devices are used to reduce symptoms or prevent injuries by correcting abnormal limb mechanics.[1,2] A new corrective orthotic termed the Flat Foot Insole™ (FFI) has been developed to correct uncompensated forefoot varus, a component of the flexible flat foot. The new orthosis is a prefabricated, semi-rigid device with a forefoot component that dorsiflexes the first two metatarsal rays to compensate for the forefoot varus. Additionally, there is longitudinal arch support in the midfoot component of the orthosis. This device has been proposed to maintain normal position, motion, and function of the entire foot. The effectiveness of the FFI™ in correcting foot biomechanics has not, however, been evaluated. Therefore, the purpose of this study was to 1) quantify foot mechanics using the FFI™, and 2) compare the effectiveness of the new orthotic in normalizing foot mechanics to an existing off the shelf orthotic and a motion control running shoe

CLINICAL SIGNIFICANCE

An orthotic device that controls forefoot varus and normalizes foot mechanics may reduce lower extremity symptom complaints and injury risk.

METHODS

Fifteen subjects (Male=4, Female=11; Mean age=34 years) with a hyper-pronated, flexible flat foot were recruited for the study. Participants were asymptomatic during daily ambulation and able to run at least one mile without pain. Subjects were not eligible for study participation if they had a history of foot surgery, peripheral vascular disease, neuropathy, or other foot abnormalities. Individuals who met all eligibility criteria and agreed to participate in the study provided written informed consent approved by the Institutional Review Board before testing procedures were initiated.

Kinematic data were recorded with a 10-camera Motion Analysis EVa RealTime system (Motion Analysis Corp., Santa Clara, CA) at 300 Hz. Twelve retro-reflective markers were placed on the right lower leg and foot[3] of each subject to track 3D movement. Windows were cut into the shoe to accommodate marker placement directly on calcaneus and forefoot landmarks. Kinetic data were collected at 600 Hz from four force plates (two each Kistler, Amherst, NY; AMTI, Watertown, MA) and synchronized with the motion data.

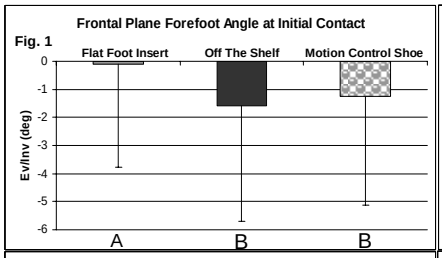
Data were collected during walking and jogging. At each speed, three trials each were collected while wearing a New Balance motion control running shoe (New Balance Athletic Shoe, Inc., Boston, MA), wearing the motion control running shoe with a semi-rigid off-the-shelf arch support insole (OTS), and wearing the motion control shoe with the FFI™. Trials were normalized to 100% of stance and trials were averaged for each subject for statistical analysis.

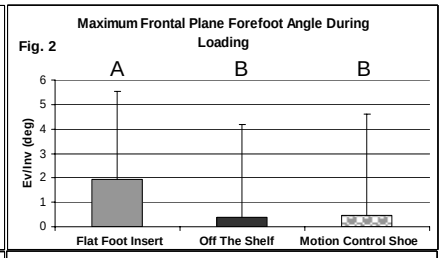
Custom Matlab programs (Matlab 7.0, MathWorks, Natick, MA) were used to calculate the 3D kinematics and kinetics for the rearfoot (calcaneus relative to tibia) and

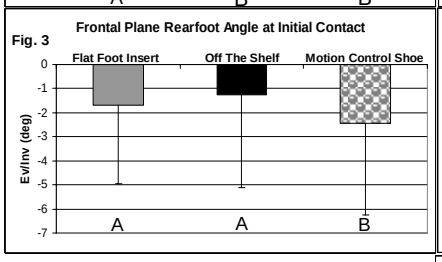
forefoot (forefoot relative to calcaneus). Statistical analysis was performed using a repeated measures ANOVA with two factors (ORTHOTIC: shoe, OTS, FFI™; CONDITION: walking, jogging). When significant main effects were identified, post hoc comparisons ($\alpha \leq .05$) were performed using Student –Newman-Keuls tests to identify the level at which differences were occurring.

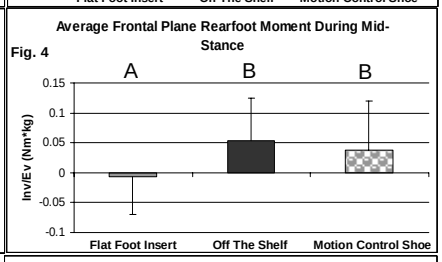
RESULTS

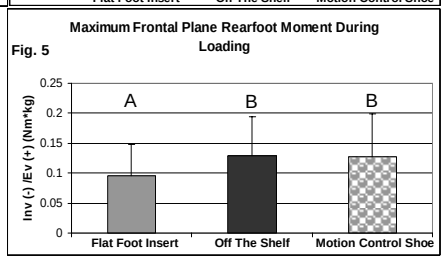
There was a trend towards significant differences for forefoot angle at initial contact (P=.056) (Fig. 1) across speeds for the three orthotic conditions. Maximum forefoot angle during loading was significantly different (P=.026) (Fig. 2). No other differences in forefoot kinematics were identified. There was a trend towards statistical significance for differences in rearfoot angle at initial contact (P=.083) (Fig. 3). No other frontal plane rearfoot kinematic differences were identified. There were statistically significant differences across speeds for orthotic conditions for both the average (P<.001) (Fig. 4) and maximum (P<.001) (Fig. 5) rearfoot moment.











SUMMARY/CONCLUSIONS

The FFI™ was effective in promoting more normal foot mechanics. This was evident as the orthosis positioned both the calcaneus and forefoot in more neutral positions, and lower external rearfoot eversion moments were recorded throughout stance. Furthermore, the addition of a traditional off the shelf orthosis while wearing a motion control shoe did not yield remarkable differences compared to wearing the motion control shoe alone.

These results support the use of this device in the clinical setting when the goal is to correct the dynamic position of a flexible flatfoot. Future studies are necessary, however, to evaluate the effectiveness of the device in alleviating pain complaints among symptomatic individuals.

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ACKNOWLEDGEMENTS

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Use of the Floor Reaction Ankle Foot Orthosis in Children with Cerebral Palsy: Efficacy and Indications Based Upon Quantitative Gait Analysis

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Introduction: The floor reaction ankle foot orthosis (FRAFO) is one type of ankle foot orthosis that is commonly prescribed for children with cerebral palsy (CP) who ambulate with excessive ankle dorsiflexion and excessive knee flexion during stance phase [1, 2]. The FRAFO is thought to be effective by limiting ankle dorsiflexion and reducing knee flexion by controlling the amount of tibial advancement over the foot during second rocker. This improved ankle and knee position is thought to reduce the elevated internal knee extensor moment commonly associated with crouch gait [2].The purpose of this study was to objectively evaluate the efficacy of the FRAFO in children with CP and to identify clinical parameters that may compromise the function of the orthosis.

Clinical Significance: There has been limited objective, quantitative research examining the effects of the FRAFO on ambulation in children with CP. Contraindications to use of the FRAFO, such as knee and hip contractures and transverse plane malalignments such as excessive external tibial torsion, have been anecdotally reported [2].

Methods: Twenty-seven children (54 limbs) with CP were identified and included in this study that had a comprehensive gait analysis performed while ambulating both barefoot and using FRAFOs in the motion analysis laboratory at our institution between January 2001 and August 2007. Gait data were collected and processed based on the Newington gait model [3]. Kinematic results included mean sagittal dynamic range of motion of the ankle in stance, peak ankle dorsiflexion, peak knee extension in midstance, and mean foot progression angle in stance. Kinetic results included minimum sagittal knee moment in midstance and were obtained only for subjects who ambulated without assistive devices. Range of motion and skeletal alignment data included knee flexion contracture, hip flexion contracture and thigh-foot angle. Repeated measures analysis of variance was used to test for mean differences between AFO and barefoot (BF) trials within subjects, with p-values of <0.05 considered significant (*). Linear correlations between variables in bivariate analyses were assessed using the Pearson correlation coefficient (r) with a significance level set at $\alpha = 0.05$ (*). Multiple linear regression analysis was used to assess possible predictive factors for peak knee extension in midstance.

Results: The mean sagittal plane dynamic range of motion of the ankle in stance went from 23° (±9°) when ambulating barefoot to 10° (±3°) with the FRAFO ($p < 0.001$*). Mean peak knee extension in midstance improved from 29° (±14°) of flexion when ambulating barefoot to 18° (±14°) when wearing the FRAFO ($p < 0.013$*). The mean minimum sagittal knee moment in midstance improved from an internal knee extensor moment of 0.2 Nm/kg (±0.3) to an internal knee flexor moment of -0.1 Nm/kg (±0.3) ($p = 0.072$). Significant correlations were found between the magnitude of knee and hip flexion contractures on physical examination and the magnitude of peak knee extension in midstance ($r = -0.784$, $p < 0.001$*, $r = -0.705$, $p < 0.001$*). Neither the clinical exam measure of thigh foot angle ($r = 0.120$, $p = 0.551$) nor the kinematic variable of mean foot progression angle in stance($r = -0.188$, $p = 0.348$) demonstrated a correlation with peak knee extension in stance with the FRAFO.

Conclusions: The FRAFO was effective in restricting sagittal plane ankle motion during stance phase. As a result, improvements in knee extension and the sagittal plane knee extensor moment in stance phase were achieved (Figure 1). This study suggests that sagittal plane limitations of range of motion, particularly knee and hip flexion contractures ≥15°, are the more significant predictors of efficacy of the FRAFO than transverse plane skeletal deformities.

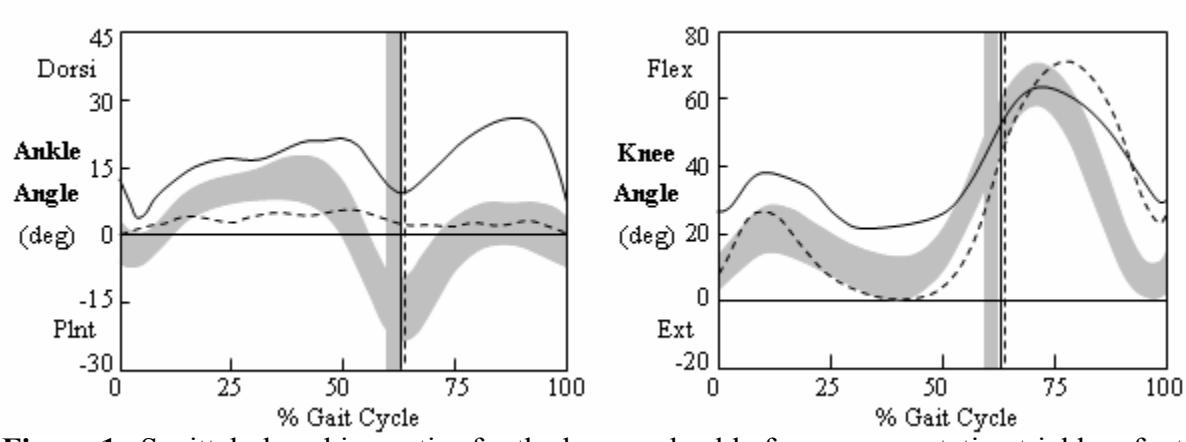

Figure 1. Sagittal plane kinematics for the knee and ankle for a representative trial barefoot (solid) versus FRAFO (dashed).

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EVALUATING MECHANICAL ANKLE FOOT ORTHOSIS CHARACTERISTICS USING BRUCE: A RELIABILITY STUDY

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Introduction

The mechanical characteristics of AFOs, such as stiffness and neutral angle around the ankle and forefoot joints, are rarely quantified. Yet, it is expected that these characteristics determine the function of the AFO in pathological gait [1,2]. Therefore, there is a need for objective measurements of the aforementioned mechanical characteristics. In this study we report on the reliability of measurements obtained with the newly developed device to determine AFO characteristics, named BRUCE.

Clinical Significance

The results derived using BRUCE can help gain insight into the role of the mechanical characteristics of AFOs in correcting pathological gait. In clinical practice, objective information of AFO characteristics is expected to lead to a better founded prescription of AFOs, resulting in more functional benefit for the patient.

Figure 1. BRUCE (The **B**i-articular **R**eciprocating **U**niversal **C**ompliance **E**stimator)

Methods

BRUCE was designed to allow manually driven flex/extension of AFOs about the ankle or forefoot joints while continuously registering joint configuration and moments exerted by the AFO onto the replicated leg [Figure 1]. From this information, neutral angles and stiffnesses around the ankle and forefoot joints are determined using a linear fit. The reliability of the stiffnesses and neutral angles was studied by repeatedly measuring the mechanical characteristics of four different AFOs. The inter-session, intra-session, and inter-observer errors were evaluated using Generalizability Theory [3].

Results

An example of the measurement results for the four different AFOs evaluated around the ankle by three different testers shows high repeatability [Figure 2]. For the four different AFOs, stiffnesses from 0.16 to 1.56 Nm/deg were found for the ankle, with neutral angles ranging from -.7 tot 6.5 deg. For the forefoot joint, stiffnesses from .09 up to .53 Nm/deg were found, with neutral angles from 3.2 to 15.8 deg. The reliability study revealed that ankle and forefoot stiffness could be measured with very high reliability, given the Standard Errors of Measurement (SEM) of .02 and .03 Nm/deg respectively. Ankle and forefoot neutral angles

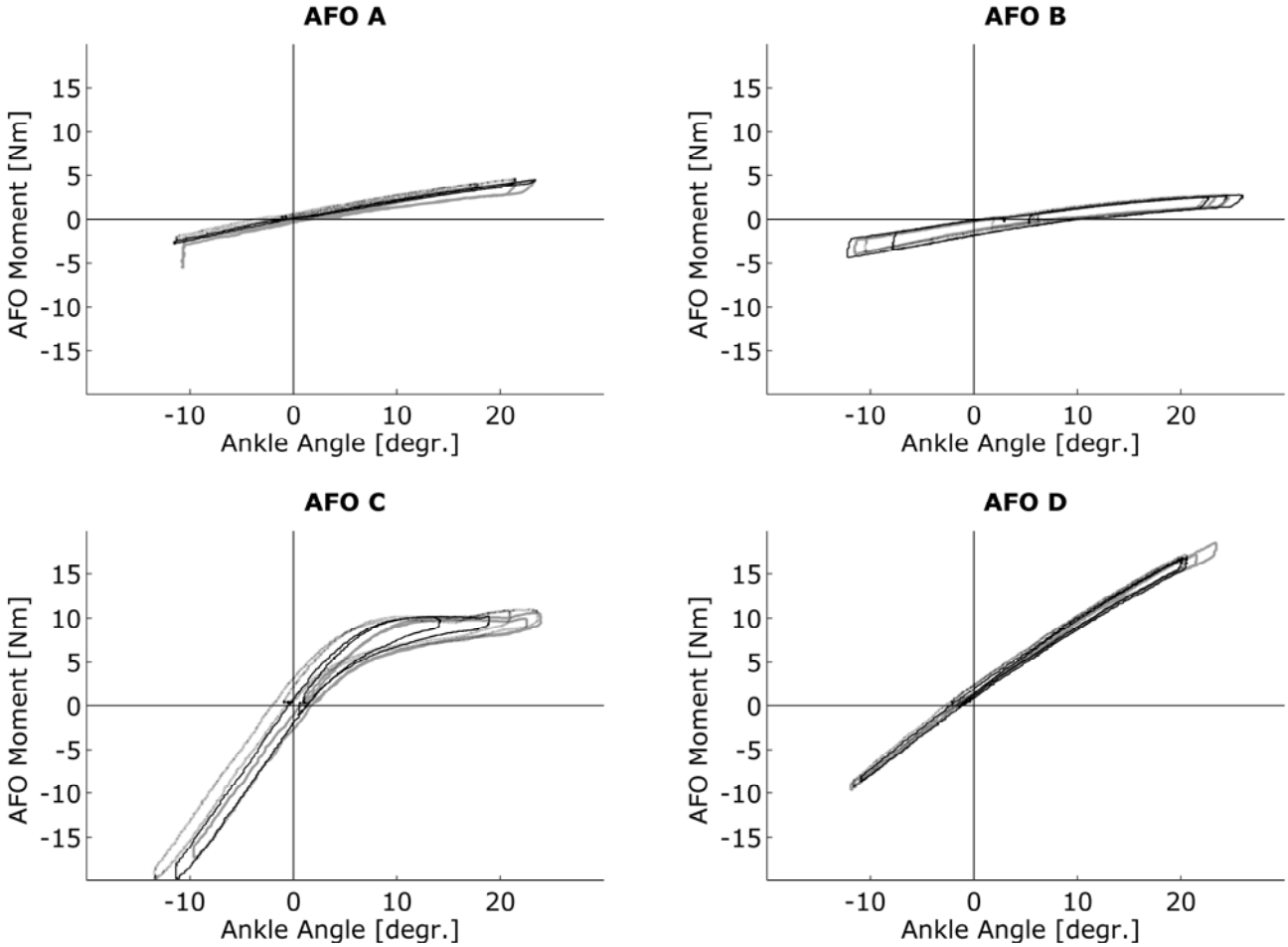

Figure 2. An example of results of the ankle, for each of the three testers, for the four tested AFOs. The results are show high repeatability.

showed SEMs of 1.1 and 2.6 deg. This measurement error could mainly be attributed to the difference in testers, since with a fixed tester SEMs of .3 and .6 deg were obtained.

Discussion

We assessed the reliability of BRUCE, the newly developed device to measure AFO characteristics The results revealed that BRUCE can reliably measure AFO characteristics such as stiffness and neutral angle around both the ankle joint and the forefoot joint. The measurement errors for the neutral ankle angle and the neutral forefoot angle could strongly be reduced by keeping the tester fixed. This might be related to the standardization of the positioning the AFO when clamping it into BRUCE. Such standardisation is important for the application of the results of BRUCE to instrumented gait analysis. It is expected that information of the mechanical characteristics of AFOs will help gain insight into the role of AFOs in correcting pathological gait. Ultimately this will result in the definition of prescription guidelines for optimal AFO-patient matching

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SHORT AND LONGER TERM CHANGES IN AMPUTEE WALKING PATTERNS DUE TO INCREASED PROSTHESIS INERTIA

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[2]Iowa State University, Ames, IA

Introduction

Effects of altering inertia properties of lower extremity prostheses on the mechanics and energy costs of amputee locomotion has been a focus in the literature [1, 2]. However, the time used to accommodate amputees to altered inertia properties has varied among studies; often accommodation periods are not reported. The accommodation process was the focus of two recent studies [3, 4] investigating non-amputees walking with additional mass added to only one leg. Both studies found asymmetrical loading produced significant gait asymmetries within the first five minutes of exposure to the load. However, it remained unclear whether these results could be extended to transtibial amputees. Thus, the purpose of this study was to quantify both short and longer term changes in the walking pattern of unilateral, transtibial amputees when the inertia properties of the prosthetic leg were matched to those of the intact leg. It was hypothesized that changes in temporal and joint kinetic variables due to alterations in prosthesis inertia would be rapid and would be complete within five minutes.

Statement of Clinical Significance

Increasing the inertia of the prosthesis exacerbated existing temporal asymmetries, but reduced existing kinetic asymmetries. Consistent with previous findings [3, 4], changes in gait symmetry occurred within the first five minutes of exposure to altered prosthesis inertia suggesting that amputees were able to accommodate to the additional prosthesis mass rapidly. Thus, changing prosthesis inertia does not require long term accommodation periods.

Methods

Four unilateral, transtibial amputees (3 males and 1 female, $M_{age} = 43 \pm 7$ yrs, $M_{height} = 1.77 \pm 0.10$ m, $M_{mass} = 102.6 \pm 4.2$ kg), completed three test sessions over eight days (Day 1, Day 2, and Day 8). On Day 1, after completing unloaded overground walking trials, the inertia properties of the prosthesis were matched to those of the intact shank and foot by adding lead shot distally to the prosthesis (mean mass added = 2.15 kg). Prosthesis inertia properties were estimated based on oscillation and reaction board techniques [1], whereas inertia properties of intact segments were estimated from regression equations [5]. Gait was reassessed after increasing inertia and then periodically over the next 60 minutes (i.e., at 0, 5, 10, 15, 30, and 60 min). Subjects then wore the additional mass during daily activities for the next 7 days. Gait assessments were completed on Day 2 and Day 8. On Day 8, gait assessments were completed first with the mass in place. The mass was then removed and gait was assessed over the next 60 minutes. Motion and ground reaction force data were collected during overground trials. Average swing and stance times were quantified from an instrumented treadmill. Peak hip and knee joint moments prior to swing initiation and termination were identified and integrals of rectified moments were obtained with respect to normalized time. Symmetry indices [1] were computed for temporal and joint kinetic variables and assessed statistically using ANOVA to determine the effect of loading on walking symmetry over time.

Results

Walking symmetry was altered systematically and immediately when prosthesis inertia was changed (e.g., Figure 1), both with the addition of mass (Day 1) and upon removal of mass (Day 8). Existing temporal asymmetries were exacerbated by increasing prosthesis inertia and were due to increased swing times of the prosthetic leg and increased stance times of the intact leg. Upon removal of the load, walking symmetry returned to baseline levels within five minutes. Only one of six joint kinetic measures changed significantly following a change in prosthesis inertia (see Figure 1).

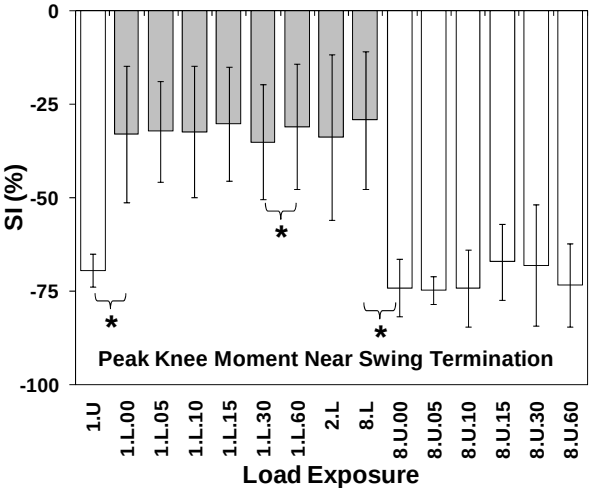

Figure 1. Symmetry indices for peak knee flexor moment near swing termination (SI = 0 indicates symmetrical gait). X-axis labels identify time period and loading condition, where the first number refers to day (1, 2, 8), the next character (U or L) refers to unloaded or loaded, and the next two characters (00-60) refer to minutes of treadmill protocol. After increasing prosthesis inertia, the prosthetic leg's knee moment increased, resulting in a more symmetrical pattern for this variable. Upon removal of the mass the asymmetry returned to baseline levels. *indicates a significant difference at $p<0.05$.

Discussion

Consistent with our hypothesis and previous results in non-amputees [3, 4], changes in stance and swing time symmetry indices were apparent in the first assessment following exposure to the altered prosthesis inertia. Changes in stance and swing time were consistent with previous results [1] which matched inertia properties between legs. Joint kinetic measures were less supportive of a change in walking symmetry following an alteration in prosthesis inertia; only one out of six measures changed significantly with changes in prosthesis inertia. The knee joint moment of the prosthetic leg near swing termination increased as inertia increased, consistent with previous results [2], whereas the intact leg's knee moment was unaltered. Thus, as inertia increased, symmetry for this variable improved. Despite improved symmetry, the higher joint moment magnitude suggests a greater muscular involvement during the swing phase, which would contribute to higher metabolic costs during walking [1].

Acknowledgements

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WALKING ON GRAVEL AFFECTS VARIABILITY AND DYNAMIC STABILITY DIFFERENTLY IN NON-IMPAIRED SUBJECTS AND TRAUMATIC AMPUTEES

Jonathan B. Dingwell[1], Shawn J. Scott[1,2], and Jason M. Wilken[2]

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[2]The Center for the Intrepid, Brooke Army Medical Center, Ft. Sam Houston, TX, USA.
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INTRODUCTION

Patients with locomotor impairments are especially challenged when walking over uneven or unstable surfaces [1]. To date, however, these studies have not yet controlled for walking speed. Also, while these studies have shown increased variability when patients walk on these surfaces, variability may not be readily equated with stability [2]. Hof [3] introduced a dynamic stability margin (DSM) based on how the center of mass (COM) moves relative to the base of support. Counter-intuitively, unilateral trans-femoral amputees exhibited *larger* DSMs on their involved sides, which also increased at faster walking speeds [4]. DSMs have not been assessed when walking stability was overtly challenged in patients. This study directly determined how both walking speed and an uneven surface affected both gait variability and DSMs in non-impaired adults and adults with unilateral trans-tibial amputation (TTA).

CLINICAL SIGNIFICANCE

Patients with amputation have a greatly increased risk of falling, which increases further in challenging environments. Understanding how these patients regulate dynamic stability when walking in complex environments is key to developing effective rehabilitation interventions.

METHODS

Seven unimpaired male military service members (age 19-29) and 10 male military service members with unilateral TTA participated. Patients wore their own regular prostheses, which varied across subjects based on individual needs. Patients walked at 3 non-dimensional speeds over both a level walkway (OG) and a loose gravel surface (RO). Subjects completed multiple trials over each surface at each speed. Full body kinematics were recorded during each trial using Motion Analysis. A 15-segment model was used to estimate whole body COM motions.

Means and standard deviations of step length, step time, and step width were calculated for each condition. Frontal plane DSMs were defined as the minimum perpendicular distance between the velocity adjusted COM vertical projection and the lateral borders of the forward supporting foot [3]. In theory, larger DSMs should reflect a safer (i.e., more stable) gait pattern [3]. Minimum DSM values during the stride were extracted for statistical analyses. A three factor ANOVA was used to determine differences between limbs, speeds, and surfaces.

RESULTS

There were no differences between limbs for either group for any measure ($p > 0.185$). Both groups exhibited greater variability when walking on the gravel (RO) surface (Fig. 1; $p < 0.005$) and greater step time variability when walking at slower speeds (Fig. 1; $p < 10^{-4}$).

DSMs exhibited minima shortly after heel strike, as in [3,4]. In non-impaired subjects, min. DSMs (Fig. 1) did not change with either walking speed ($p = 0.298$) or surface ($p = 0.773$). Amputees actually exhibited *larger* min. DSMs than non-impaired (Fig. 1), similar to [4]. In amputees, min. DSMs were *greater* when patients walked on the gravel surface ($p = 0.039$) and *increased* further when these patients walked at faster speeds ($p < 10^{-4}$).

There were occasional individual subject trials over the loose gravel where the DSMs were only a few millimeters. During these trials, some subjects exhibited stumbling responses, including occasional cross-over stepping. However, the DSMs never became negative and no subject experienced a fall.

SUMMARY/CONCLUSIONS

Patients adapted to the unstable gravel surface by *increasing* their DSMs, similar to [4]. This was largely due to taking wider steps. This provides additional evidence that DSMs reflect how patients *compensate* to maintain balance when their stability is challenged. Therefore, they may *not* reflect a direct measure of fall risk *itself*. It is not yet known if these changes in DSM directly predict changes in fall risk.

The lack of differences between intact and prosthetic limbs in TTA patients contrasts with results in trans-femoral patients [4]. This was likely due either to differences in level of amputation, activity/fitness level, or both. Further research needs to determine how gait symmetry, stability, and fall risk are inter-related. The present findings provide important insights into how TTA patients adapt their gait patterns to challenging walking environments.

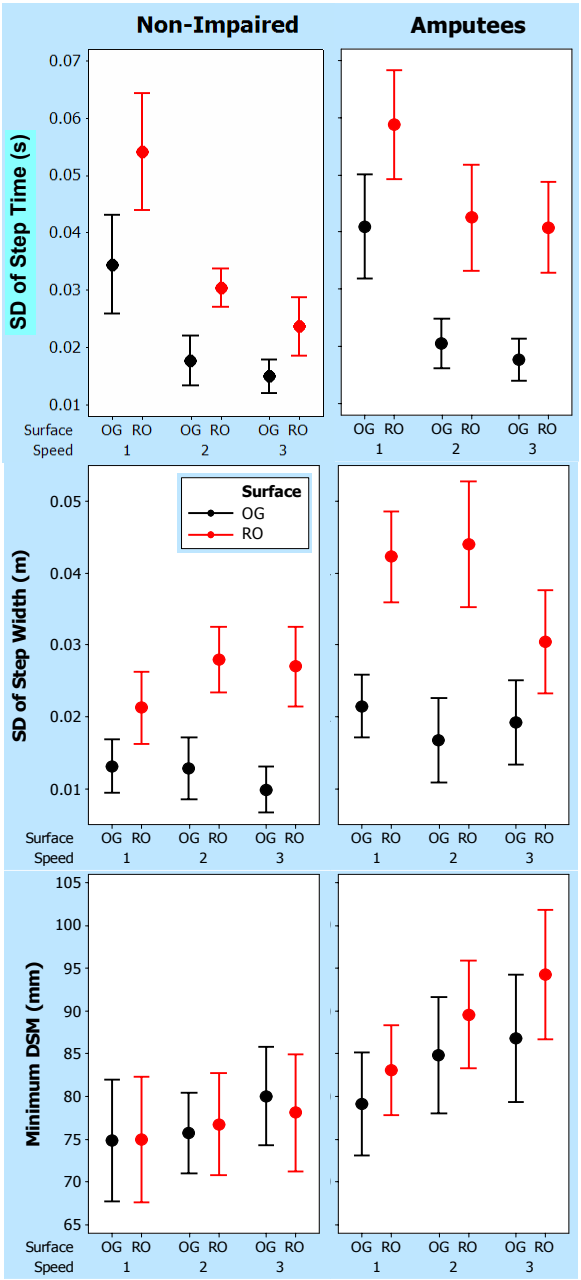

Figure 1: Variability of stride time (top) and step width (middle) and minimum dynamic stability margins (bottom) for non-impaired (left) and amputee (right) subjects as a function of walking speed and surface ("OG" = overground, "RO" = Gravel rocks). Error bars indicate ± 95% CI.

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MEDIOLATERAL GROUND REACTION IMPULSES, STRIDE LENGTH AND EFFECTIVE LIMB LENGTH OF AMPUTEES TURNING VS. NONAMPUTEES

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INTRODUCTION

Turning is required for successful household and community ambulation[1], comprising about 20% of all daily steps[2]. Altered mediolateral ground reaction impulses (ML GRI) are thought to be the primary mechanism non-amputees use to change heading, while changes in limb length resist the centripetal forces created with circular motion[3]. Transtibial amputees may compensate for their limited range of motion and control at the ankle by using different turning strategies. The purpose of this study was to compare ML GRI and a measure of effective limb length and stride length to identify if transtibial amputees have a different turning strategy compared to non-amputees.

CLINICAL SIGNIFICANCE

Understanding the biomechanical differences between amputees and non-amputees during turning may serve to identify where amputees compensate, leading to prosthetic interventions designed to minimize these compensations that might lead to secondary injuries.

METHODS

Subjects: Nine transtibial amputees (55±13 years, 1.8±0.1m ht, and 89±11kg wt) and nine non-amputees (42±17 years, 1.69±0.1m ht, and 70±15kg wt) gave informed consent to participate in this IRB-approved study. ***Procedures:*** Subjects walked clockwise and counter-clockwise along a 1m-radius circular path at self-selected turning walking speed. Gait biomechanics were assessed using a 12-camera Vicon system and Plug in Gait model (Centennial, CO) and two embedded force plates (Bertec, Columbus, OH). Mediolateral ground reaction force impulses (ML GRI, integral of ML force during stance), stride length difference (outside – inside limb), and peak effective limb length difference (resultant distance between the hip joint center and center of pressure during single limb stance; outside - inside) were calculated for non-amputees (NA), amputee prosthetic limb (PR) and sound limb (SO) and averaged across trials for each condition. The inside and outside limbs were analyzed separately since previous literature reported differences in turning limb for non-amputees[3]. Differences across conditions were determined using linear mixed effects models with the limb type (NA, PR, and SO) as the fixed effect and subject as the random effect.

RESULTS

ML GRI, stride length and peak effective limb length were calculated for each condition (Table 1). ML GRI and stride length decreased for PR and SO on the inside of the turn compared to NA. The effective limb length for SO on the outside of the turn was found to be greater than PR and NA. A larger difference in stride length was found for SO (as the outside limb) compared to NA and PR (Table 2). A larger difference in effective limb length was found for PR (as the outside limb) compared to NA. A slightly faster speed was

observed for NA compared to the amputees (0.95 ± 0.1 vs. 0.86 ± 0.1 m/s), but the difference between speeds was within one standard deviation.

Table 1: ML GRI, stride length and effective limb length for inside and outside limbs turning. Pair-wise comparisons made between conditions with statistical significance set at $p < 0.05$ and NS indicates $p>0.1$.

	Non-amputee	Prosthetic	Sound	*p* (PR v NA)	*p* (SO v NA)	*p* (SO v PR)
ML GRI (Nsec/kg)						
Inside	0.3 ± 0.16	0.006 ± 0.13	0.057 ± 0.16	**0.0001**	**0.0013**	NS
Outside	0.71 ± 0.15	0.78 ± 0.09	0.79 ± 0.11	NS	NS	NS
Stride length (mm)						
Inside	1061 ± 64	952 ± 82	991 ± 112	**0.0025**	0.086	NS
Outside	1234 ± 92	1212 ± 129	1205 ± 123	NS	NS	NS
Eff. Limb length (mm)						
Inside	866 ± 73	935 ± 61	938 ± 56	NS	NS	NS
Outside	867 ± 70	923 ± 58	932 ± 58	0.076	**0.041**	**0.0004**

Table 2: Differences (outside-inside) between stride length and effective limb length for NA compared to PR and SO (outside limb). Statistical significance was set at $p < 0.05$ and NS indicates $p>0.01$.

DIFFERENCES	Non-amputee	Prosthetic	Sound	*p* (PR v NA)	*p* (SO v NA)	*p* (SO v PR)
Stride length (mm)	172 ± 71	221 ± 39	253 ± 54	0.068	**0.01**	**0.02**
Eff. limb length (mm)	1.1 ± 10.2	-11.1 ± 13.8	-3.3 ± 19.1	**0.023**	NS	0.057

DISCUSSION

This study demonstrated decreased ML GRI on the inside limb for both prosthetic (PR) and sound (SO) compared to non-amputee (NA). The amputee may have decreased ML GRI on the inside limb to provide less resistance in the direction of the turn, thereby facilitating the turn. This decreased resistance may be necessary if the amputees walk with a wider step width to maximize their base of support and consequently stability. The amputee also demonstrated a larger difference between the outside and inside limb stride length as compared with NA, which was due to a smaller stride on the inside limb. Finally, the NA outside effective limb length was slightly longer than the inside limb, resulting in a positive difference (outside – inside). Although SO outside effective limb length was shorter than the inside limb (negative difference), it was not statistically different from NA. However, PR had a larger negative difference which was statistically significant compared to NA and borderline significant compared to SO. A longer effective limb length on the inside of the turn was unexpected since leaning into the turn would be expected to cause a longer effective outside limb length. Other metrics such as pelvic obliquity, center of mass trajectory, knee angle and step width should be examined for further interpretation of these results. A prosthetic device that minimizes these biomechanical differences may lead to enhanced amputee performance during daily activities and a decrease in secondary injury.

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LOWER EXTREMITY ROTATION MOMENTS FOR AMPUTEES TURNING COMPARED TO NONAMPUTEES

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INTRODUCTION

Amputees must compensate for their lack of movement and control mechanisms at the foot and ankle in order to ambulate. Turning, which is associated with navigating in home and community environments[1], may require more complex biomechanical compensations for amputees compared to non-amputees. However, the specific adaptations for amputee maneuvering have not been clearly defined. Because non-amputees change their body orientation during a turn by primarily applying transverse plane rotation moments to the ground[2], the purpose of this study was to identify if amputees use a similar strategy to change orientation during a turn.

CLINICAL SIGNIFICANCE

Understanding the biomechanical differences between amputees and non-amputees during turning may serve to identify where amputees compensate, leading to prosthetic interventions designed to minimize these compensations that might lead to secondary injuries.

METHODS

Subjects: Ten transtibial amputees (56±12 years, 1.8±0.1m ht, and 88±11kg wt) and ten non-amputees (43±15 years, 1.7±0.1m ht, and 75±23kg wt) gave informed consent to participate in this IRB approved study. ***Procedures:*** Subjects walked clockwise and counter-clockwise along a 1m-radius circular path at self-selected turning walking speed. Gait biomechanics were collected using a 12-camera Vicon system and Plug in Gait model (Centennial, CO) and two embedded force plates (Bertec, Columbus, OH). Peak internal and external rotation moments at the hip, knee and ankle were extracted for non-amputees (NA), amputee prosthetic limb (PR) and sound limb (SO) using the Event Analyzer (Vaquita Software, Zaragoza, Spain) and then averaged across trials (3 repeated trials) for each condition. The inside and outside limbs were analyzed separately since previous literature reported differences in turning limb for non-amputees[2]. Differences across conditions were determined using linear mixed effects models with limb type (NA, PR, SO) as the fixed effect and subject as the random effect. Statistical significance was set at $p < 0.05$.

RESULTS

Peak hip and knee rotation moments for NA, PR and SO conditions are presented in Table 1. Differences across condition are portrayed in figure 1 for the hip and knee rotation moments for the inside and outside limbs, respectively. Peak internal knee and hip rotation moments were decreased for PR compared to NA for both the inside and outside limbs during a turn. PR was also decreased compared to SO for the knee internal rotation moment, with a similar trend and borderline significance at the hip. There were no significant differences in ankle rotation moment, although there was a similar trend for the internal rotation moment: PR was

less than SO and NA. Amputees and non-amputees walked at comparable speeds (0.88 ± 0.1 m/s and 0.93 ± 0.1 m/s, respectively).

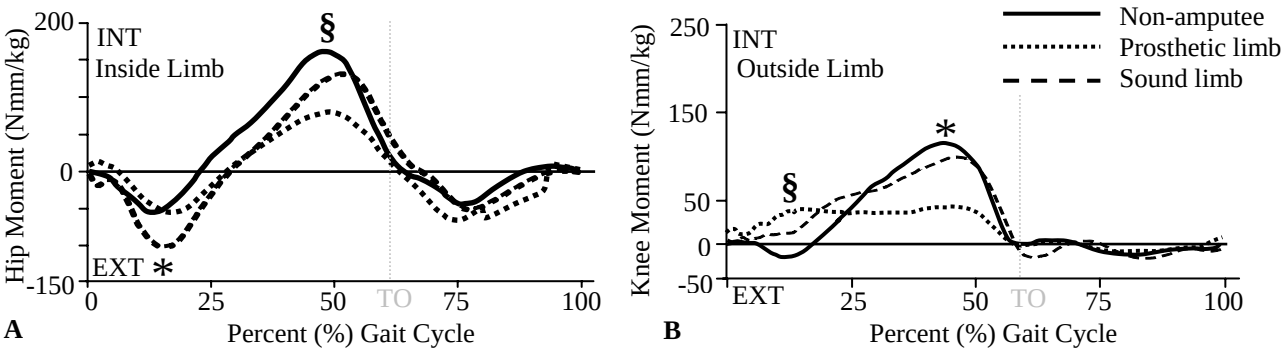

Figure 1: (A) Hip (inside limb) and (B) knee (outside limb) rotation moments across the gait cycle. § indicates prosthetic limb is less than non-amputee. * indicates sound limb was greater than prosthetic and non-amputee for the hip and prosthetic limb is less than sound and non-amputee for the knee.

Table 1: Peak internal and external knee and hip rotation moments for the inside and outside limbs during a turn for non-amputee (NA), prosthetic (PR) and sound (SO) conditions. Statistical significance was set at *p* < 0.05 and NS indicates *p* > 0.1.

	Non-amputee	Prosthetic	Sound	*p* (PR v NA)	*p* (SO v NA)	*p* (SO v PR)
Knee (Nmm/kg)						
Inside internal	206 ± 36	157 ± 43	216 ± 61	**0.0027**	NS	**0.02**
Outside internal	122 ± 37	70 ± 26	121 ± 67	**<.0001**	NS	**0.048**
Inside external	-30 ± 20	-26 ± 27	-34 ± 34	NS	NS	NS
Outside external	-28 ± 24	-7 ± 21	-16 ± 18	**0.0043**	0.056	0.094
Hip (Nmm/kg)						
Inside internal	174 ± 43	97 ± 42	148 ± 87	**<.0001**	NS	0.07
Outside internal	124 ± 47	53 ± 36	88 ± 61	**<.0001**	0.091	0.054
Inside external	-66 ± 38	-54 ± 47	-129 ± 62	NS	**0.0066**	**0.0077**
Outside external	-102 ± 60	-70 ± 54	-89 ± 48	NS	NS	NS

DISCUSSION

This study demonstrated differences in rotation moments for transtibial amputees turning compared to non-amputees. The increased peak external hip rotation moment for inside SO as compared to PR and NA (figure 1A) maybe required to compensate for the decreased peak internal rotation moment at all joints for PR compared to SO and NA. This decreased internal rotation moment for PR may be a protective mechanism the amputees use to minimize torques on the residual limb. A prosthetic device that has adjustable transverse plane rotation may facilitate amputee turning by allowing for rotation without excessive torque at the residual limb-socket interface.

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GASTROCNEMIUS MUSCLE TENDON UNIT LENGTH: CLINICAL ASSESSMENT VERSUS KINEMATIC ASSESSMENT IN DIABETIC SUBJECTS

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Introduction

Individuals with diabetes mellitus (DM) often develop muscle stiffness that may lead to equinus and has been hypothesized to increase pressure on the plantar surface of the forefoot during gait. Surgical plantarflexor lengthening is sometimes performed in an effort to reduce forefoot pressures (FP) and ameliorate the risk of ulceration[1-4]. The decision to perform surgery is often dependent on clinical assessment of equinus with the patient seated and the knee fully extended. However, there has been no prior study of the relationship between gastrocnemius muscle-tendon unit (MTU) length measured during this passive clinical ankle dorsiflexion assessment (PCADA) and measured functionally during standing and gait. The gastrocnemius is biarticular, and thus its MTU length depends both on knee and ankle angle. It is therefore possible that patients with clinically measured equinus may choose to walk with a compensatory strategy such as mild knee flexion and thus never reach the gastrocnemius MTU length achieved during PCADA. The purpose of this investigation was to understand the relationship between gastrocnemius MTU length measured during PCADA and maximum MTU length measured during standing and walking in both diabetic and control subjects.

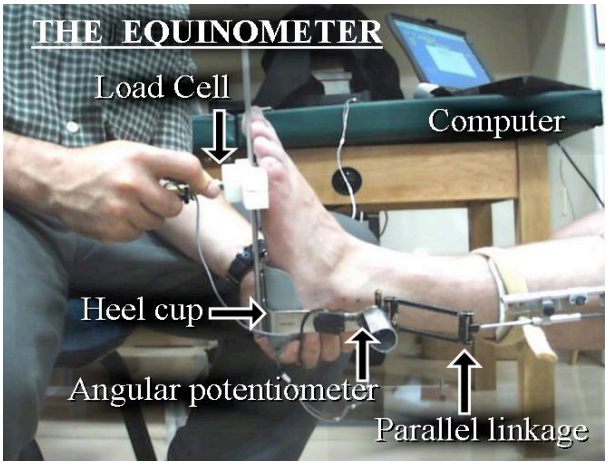
Statement of Clinical Significance

Knowledge of the relationship between PCADA and gastrocnemius MTU length during standing and walking will enhance understanding of the contributions of MTU length to FP, and therefore understanding of the potential benefits of surgical lengthening procedures as a tool for modifying FP.

Methods

Ten individuals with DM (Age 57.1 ± 6.2; Wt 97.7 ± 23.4 kg, Ht 172 ± 7 cm; duration of diabetes 12.2 ± 3.8 yrs) and 10 healthy control subjects (Age 56.3 ± 10.3; Wt 86.7 ± 17.2 kg, Ht 173 ± 8 cm) provided informed consent to participate in this IRB-approved protocol. PCADA dorsiflexion angle was assessed using a device called the equinometer[5] (see figure 1) that has been validated in prior study[6]. This device uses a load cell and an angular potentiometer mounted onto a small frame that holds the foot and shank. Similar to a standard clinical ankle exam[5], the subject was seated with a fully extended knee while the ankle was dorsiflexed with 10 ± 1 Nm of

Figure 1. The equinometer.



torque for 5 seconds. Dorsiflexion was recorded from the angular potentiometer and averaged over 3 trials.

Gait data was then collected using a 12-camera Vicon 612 system and the Full Body Plug in Gait model (Oxford Metrics, Oxford, UK). Subjects completed 1 static standing trial and 5 trials walking barefoot at 0.9 m/s ± 10% straight ahead along an 8m walkway. 3D kinematics were calculated with Nexus software and all MTU lengths were calculated utilizing a model based on anatomic MRI data reported by Eames et al[7] with inputs of ankle and knee angle and shank length. MTU length maximums were recorded during the static standing trial, during the PCADA, and during the dynamic gait trials. Linear mixed effects models were used to determine if there was a difference in MTU length between conditions (PCADA, standing, walking). Significance was set at $p < 0.05$ *a priori.*

Results

There were no significant differences between the diabetic group and the control group in terms of age (p= 1.0), BMI (p=0.3) or gastrocnemius MTU length (p=0.8).

In the control group, gastrocnemius MTU length was found to be essentially the same during PCADA and at its peak during barefoot standing and walking. In the diabetic group, although not statistically significant, there was a trend toward peak MTU length during barefoot standing and walking exceeding MTU length by 2-3 mm during PCADA (p=.087).

Discussion

This study demonstrated that in a population of diabetic and control subjects with an average ankle dorsiflexion limit of motion of 12.0 and 11.4 degrees respectively, the peak functional gastrocnemius MTU length during standing and walking did not significantly exceed the gastrocnemius MTU length measured during PCADA. This indicates that the relative effect of passive tension on FP is unlikely to be clinically significant. Since plantarflexor lengthening procedures aim to reduce passive tension in the MTU by lengthening the achilles tendon, it is therefore unlikely that the desired effect of decreasing FP would be achieved in a similar population. A potential limitation of this study is the relatively small percentage of subjects with equinus (defined as <5° of dorsiflexion; only 1 out of 10 diabetic subjects and 3 out of 10 control subjects in our study). It is unclear whether a population with equinus would demonstrate increased MTU length during walking compared to during PCADA, or whether they would compensate with more knee flexion during gait therefore shortening their MTU length. Thus, further study is warranted.

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EFFICACY OF BOTULINUM TOXINE A TREATMENT IN CHILDREN WITH CEREBRAL PALSY IS DEFINED BY CRUCIAL FACTORS WITHIN THE TREATMENT STRATEGY

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INTRODUCTION

From an overall view of the literature, treatment with botulinum toxine A (BTX-A) can be considered as being successful in children with cerebral palsy (CP). However, there is still considerable variability in the amount of response to the treatment between and within patients. The influence of BTX-A injections on lower limb function is not entirely predictable so far. The aim of this study was to evaluate the clinical responsiveness of lower limb BTX-A treatment in CP and to delineate crucial factors within the treatment strategy which may predict the outcome.

CLINICAL SIGNIFICANCE

Knowledge of crucial factors that define the outcome of BTX-A injections in CP, might help the clinician to select patients most likely to benefit from the treatment, to assure the most optimal treatment strategy, and to improve the prognosis the outcome.

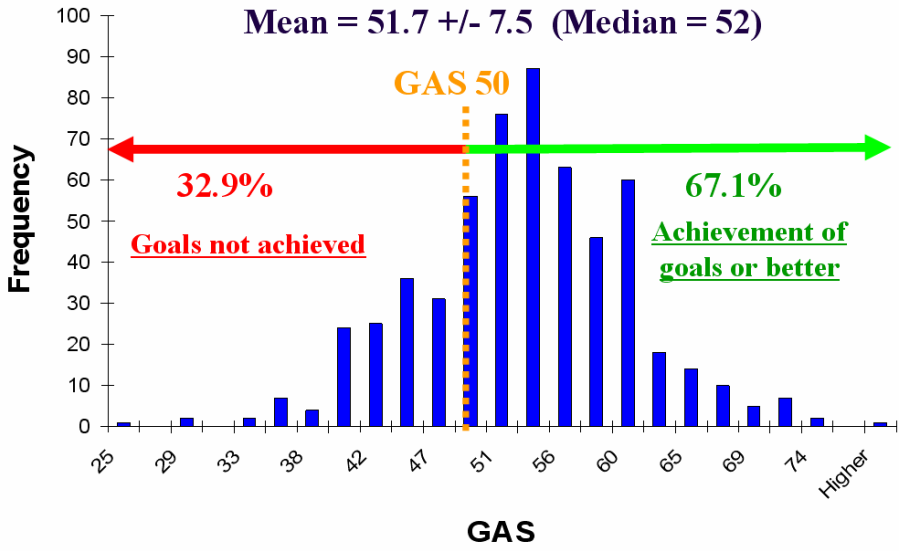
PATIENTS/MATERIALS and METHODS

From a database of all BTX-A treatments (N=2276) at the multidisciplinary clinic of CP between January 1996 and December 2006, 1090 treatments were selected based on following criteria: diagnosis of CP, age at time of treatment < 24 years, lower limb treatment, no surgery at time of BTX-A injections, standardized gait and clinical evaluation pre and 2 months post BTX-A injections. In this study, the last treatment for each patient was included for prior analysis, resulting in a final set of 577 treatments. The Goal Attainment Score (GAS) of each treatment was calculated to evaluate the efficacy. At the time of the treatment, the multidisciplinary team stipulated general goals based on all available evaluation data. General goals were then retrospectively translated into 3 to 6 specific goals (list of goals defined by CP team using gait analysis and clinical examination data). The achievement of goals was assessed by comparison of gait analysis data and clinical examination pre and 2 months post injections (according to predefined criteria for success), defined by an independent evaluator. The composite change score (based on 3 to 6 goals) was expressed as a T-score with a mean equal to 50, indicating that the expected level of responsiveness was attained. The interobserver reliability for this application of the GAS, defined in an independent study on 30 treatments, resulted in an ICC of 0.945 and SEM of 1.992. From the electronic patient files, a set of patient characteristics, injection characteristics and aftercare issues were defined. Differences of group characteristics for continuous and categorical data were analyzed by ANOVA analysis of variance (post-hoc

Tuckey) and Kruskal-Wallis test. A set of influencing variables (for which the p-value for analysis of variance was < 0.20) were subsequently combined in a multiple regression model.

RESULTS

The mean GAS for the total group was 51.7 (+/-7.5), indicating that, on average, the patients achieved the treatment goals. Considering a converted total score of 50 as cut-off for a successful treatment, 67.1% of the treatments were successful. GAS were significantly higher for the GMFCS type I compared to type III (p=0.021) and for children with hemiplegia compared to di/quadriplegia (p<0.0001). GAS significantly correlated with age at time of treatment (r=-0.23), but not with dosage. The GAS was found to be significantly higher for multilevel injections or injections only in the distal muscle groups, compared to injections only in the proximal muscles of the lower limb (p<0.0001). Other crucial factors for a successful outcome were number os previous sessions of BTX-A treatments (p<0.001) duration of physical therapy per week (p=0.0026), post injection casting (p=0.005) and frequency of using day and night orthoses after injections (p<0.0001). Multiple regression analysis revealed that five predictors (age, diagnosis, muscle selection, and day and night orthoses) made a statistically significant contribution to the prediction of the outcome. R^2 was 0.17 indicating that 17% of the treatment outcome can be explained by these four factors.



DISCUSSION

The results of this study confirm the effectiveness of BTX-A treatment of lower limb spasticity in children with CP. This study has highlighted several factors that might help the clinician to select patients most likely to benefit from the treatment and to assure the most optimal treatment strategy. BTX-A injections should not be considered as an isolated treatment. The results indicated that, apart from treatment specifications, also aftercare issues, such as casting, physical therapy and orthotic management, significantly influenced on the outcome. Based on the current results, future research should focus on the success of treatment in more involved children (GMFCS>3) and on the success of repeated BTX-A sessions.

Explicit Kinematic Criteria Predict Postoperative Improvements Following Surgical Lengthening of the Psoas Tendon in Patients with Cerebral Palsy

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INTRODUCTION

Intramuscular psoas lengthening has been shown to improve dynamic hip function when included as part of a single event multi-level surgery (SEMLS) for children with cerebral palsy [1, 2]. However, in previous analyses, there have been no control subjects to which outcomes could be compared [1, 2]. The procedure remains somewhat controversial due to a lack of clear indications for patient selection, as well as concerns about postoperative weakness and safety of the procedure. This study reports on a large retrospective cohort of subjects who underwent SEMLS treatment, with or without psoas lengthening, in order to determine if appropriate patient selection can optimize functional outcomes.

CLINICAL SIGNIFICANCE

Explicit indications for performing psoas surgery remain controversial. Identification of hip flexion contracture by physical examination is highly variable and may not identify all patients with dynamic hip flexor pathology. This study proposes and tests explicit kinematic criteria that are predictive of postoperative improvements with surgical lengthening of the psoas tendon.

METHODS

A retrospective analysis was performed on children with cerebral palsy, ages 3 to 18, who underwent SEMLS (± intramuscular psoas lengthening) at Gillette Childrens Specialty Healthcare after January 1994. Each child had an initial gait analysis no more than 18 months before the SEMLS and a follow up gait analysis between 6 and 24 months after the SEMLS. Average kinematics were computed for each side for each child and treated as a separate case. SEMLS was defined as 2 or more major surgical procedures for any one side (*e.g*. osteotomy, muscle lengthening, muscle transfer). Subjects who underwent selective dorsal rhizotomy were excluded from the study. Children were further categorized by whether they had undergone previous major surgery.

Children meeting two-out-of-three of the following criteria were considered appropriate for psoas surgery: 1) maximum hip extension greater than 8°, 2) maximum pelvic tilt greater than 24°, and 3) pelvic tilt range of motion greater than 8°. These levels represent two standard deviations from normal values. To quantify the outcome of the intervention, the change in overall gait pattern, as measured by the gait deviation index (GDI), was calculated [3]. In addition, the change in pelvic and hip kinematics was measured using a modified version of the GDI (the pelvis and hip deviation index or PHiDI) that only includes sagittal plane pelvis and hip kinematics. The PHiDI scores can be interpreted in the same manner as the GDI scores; a score of ≥100 means normal kinematics, and each 10 points below 100 is one standard deviation away from normal kinematics.

RESULTS

The inclusion criteria produced 412 sides, 211 of which had no prior surgery. These 211 sides will be the focus of the present analysis. Of these 211, 55 went on to have a psoas lengthening as part of their SEMLS treatment. Measures of kinematic function before and after intervention are shown [Table 1].

	Met Kinematic Criteria	Psoas Lengthening Included	
		Yes	No
PHiDI pre	Yes	74	77
	No	87	90
PHiDI post	Yes	84	84
	No	89	91
GDI pre	Yes	62	61
	No	69	71
GDI post	Yes	75	72
	No	76	80

Table 1 Mean GDI and PHiDI, pre and post intervention.

The most dramatic improvement in hip function (as measured by PHiDI) occurred in the group that met the kinematic criteria and went on to have psoas lengthening [Figure 1a]. This was a larger improvement than the group that met the criteria but did not have surgery, although this latter comparison was not statistically significant. The PHiDI was improved (ΔPhiDI > 0) for all subjects who met the kinematic criteria ($p < 0.001$) and was statistically unchanged for those who did not ($p > 0.15$), regardless of whether they underwent psoas lengthening [Figure 1a].

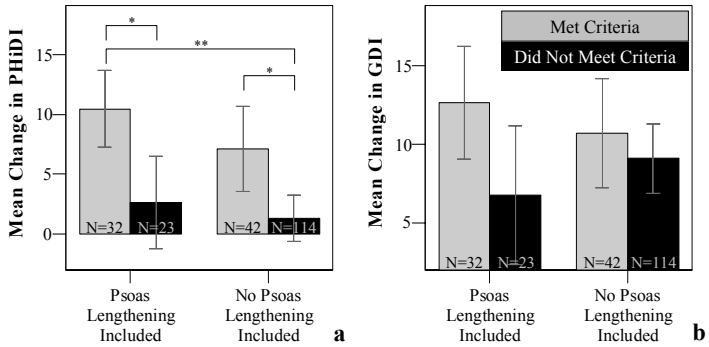

Figure 1 Change in a) PHiDI and b) GDI after a SEMLS procedure. Error bars show the 95% confidence interval. The largest changes occurred for the group that met the criteria and had a psoas lengthening. (* = $p < 0.05$, ** = $p < 0.001$)

The GDI improved after surgery (ΔGDI > 0, $p < 0.01$) regardless of whether psoas lengthening was included in the SEMLS, or whether the side met the kinematic criteria for psoas lengthening [Figure 1b]. The greatest improvement was for the group that met the kinematic criteria and had psoas lengthening, while the group with the least improvement met the criteria but did not have the surgery. There was no significant difference between any of the groups for mean change in GDI.

DISCUSSION

Overall kinematics improved for children undergoing SEMLS treatment. When deciding whether or not to include a psoas lengthening in the procedure, pelvis and hip kinematics in the sagittal plane can give guidance. Subjects who met the kinematic criteria for psoas surgery exhibited significantly more improvement in pelvis and hip kinematics than those who did not meet the criteria [Figure 1a]. The overall and hip-specific results imply that, for children who meet the kinematic criteria, the best results are realized when psoas lengthening is included as part of a SEMLS procedure. The converse is also true; children who do not meet the kinematic criteria tend to achieve the best results when psoas lengthening is left out of the SEMLS procedure. Further study will focus on examining safety, as well as investigating whether there are other criteria that can more definitively guide the decision making process.

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DISTAL TIBIAL ROTATION OSTEOTOMIES NORMALIZE FRONTAL PLANE KNEE MOMENTS

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INTRODUCTION

Excessive internal or external torsion of the tibia may potentially alter gait parameters in all planes at all lower extremity joints. While the documentation and pathomechanics of tibial torsion and outcomes of corrective osteotomies have been studied in populations such as cerebral palsy [1,2] and myelomeningocele [3], the idiopathic population has been less researched. The cerebral palsy and myelomeningocele populations often have concomitant surgeries to correct soft tissue and other bony deformities making it difficult to isolate the effects to tibial rotation surgery. In addition, knee kinetics were not reported [2] or only internal rotational osteotomies were studied [1,3]. In an idiopathic population, the effects of excessive internal or external tibial rotation on knee biomechanics can be isolated. The goal of this study is to document the effects of pathologic tibial torsion on the knee moments during gait in an idiopathic population. We hypothesized that based on the biomechanics this deformity should create abnormal frontal plane moments about the knee. This study investigated these moments in two groups of patients, one with internal torsion and the other with external torsion. Frontal plane knee moments were evaluated both prior to correction and after distal tibial rotational osteotomy to evaluate the effect of the surgery on this parameter.

CLINICAL SIGNIFICANCE

Distal rotational osteotomy of the tibia is a major bony procedure. For surgeons, patients and families to invest in this procedure, a complete understanding of the expected outcomes is necessary.

METHODS

IRB approval at two Shriners Hospitals for Children sites was obtained for this study. Inclusion criteria were limited to subjects with idiopathic tibial torsion between the ages of 8 and 18 identified as candidates for tibial rotational osteotomies. Eleven subjects were enrolled for prospective analysis over a three year period. Due to this small sample size, seven additional children were added retrospectively. These eighteen children were divided into two groups according to their deformity; the internal torsion group was comprised of eight children (n=11 limbs) and the external torsion group was comprised of ten children (n=14 limbs). All children ambulated without assistive devices (GMFCS level I). Standard gait analysis techniques were applied for both pre- and postoperative analyses to determine kinematics and kinetics of the lower limbs. Both peak and average frontal knee moments during stance were compared using paired student t-tests with α=0.05. An internal or reaction joint moment convention is used representing the moments required to balance the external load.

RESULTS

Children in the internal torsion group exhibited a preoperative internal mean foot progression angle during stance of 16°. In the external group, this value was -27°. With surgical correction

the internal group improved to 1° (p<0.001) and the external group improved to -13° (p<0.001) with normal values (mean ± standard deviation) for this parameter -9° ± 7°. Average knee valgus moments (Fig. 1) during stance significantly reduced in the internal torsion group from 0.27 to 0.15 *Nm/kg* (p=0.004), and significantly increased in the external torsion group from 0.08 to 0.19 *Nm/kg* (p<0.001). Normal values for this parameter are 0.16 ± 0.10 *Nm/kg*. Peak valgus moment in the internal torsion group significantly reduced from 0.47 to 0.36 (p=0.027) *Nm/kg* and in the external torsion group significantly increased from 0.30 to 0.41 *Nm/kg* (p=0.004). Normal values for this parameter are 0.33 ± 0.15 *Nm/kg*.

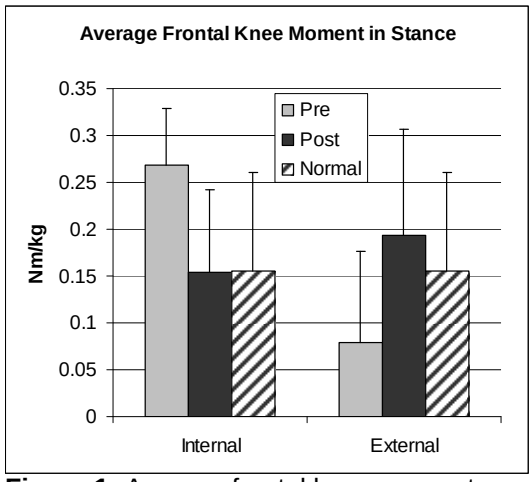

Figure 1: Average frontal knee moments during stance normalized in both internal and external subject groups following distal tibial rotational osteotomy.

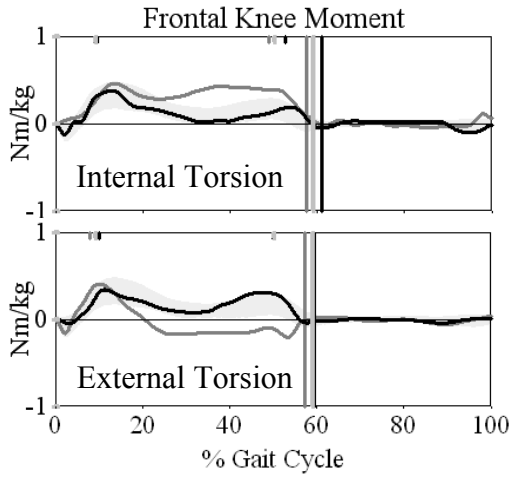

Figure 2: Frontal knee moments during gait of internal (top) and external (bottom) sample subjects. Grey = Pre, Black = Post, Light Grey Shade = Normal ± 1 Standard Deviation.

DISCUSSION:

Abnormally increased average knee valgus moment was present in subjects with internal tibial torsion. Abnormally reduced average knee valgus moment, or relative increased varus moment was present in subjects with external tibial torsion. In both cases values were corrected to within normal range following corrective surgery. Peak values also exhibited significant changes, but both groups started at values within one standard deviation of normal. From this study we conclude that abnormal frontal knee moments are present throughout stance in subjects with excessive tibial torsion. While these deformities may not alter peak values to clinically significant levels, abnormal values are present throughout stance (Fig. 2). Such moments may adversely affect the articular cartilage of the knee by overloading the medial compartment in internal torsion subjects and the lateral compartment in external torsion subjects. Surgical correction by tibial rotational osteotomy corrected these moments to within normal values. While this study was intentionally performed in subjects with idiopathic tibial torsion to eliminate or reduce confounding variables, the findings may also be applied to the more common presentation of torsional deformities found in the cerebral palsy and myelomeningocele population.

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LONG-TERM OUTCOMES OF ORTHOPEDIC SURGERY IN PERSONS WITH CEREBRAL PALSY: FEMORAL DEROTATION OSTEOTOMIES

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Introduction

Orthopedic surgery is a routine component in the treatment of gait abnormalities for children with cerebral palsy (CP). Typical surgery involves simultaneous procedures to lengthen or transfer muscles at multiple levels and correct abnormal bony alignment of the lower extremities. Specifically, in persons with excessive internal rotation of the hips and increased femoral anteversion, femoral derotation osteotomies (FDO) are performed. The 5-year outcomes of FDO have been studied[1], but the longer-term (10 year) effects of FDO on the ability of children with CP to walk are unknown. The aim of this study was to evaluate the long-term effects of FDO on gait in subjects with CP. We hypothesized that subjects who demonstrated increased palpated femoral anteversion and internal hip rotation and internal foot progression gait pattern pre-operatively would demonstrate decreased internal hip rotation and a more external foot progression angle post-operatively and would maintain these improvements over time.

Statement of Clinical Significance

Information gained from this study provides a better understanding of the long-term effects of FDO on gait in subjects with CP.

Methods

Forty subjects with CP volunteered to participate in this repeated measures design. A subset of 17 subjects (25 sides) underwent FDO in combination with soft tissue procedures and/or bony procedures to the foot. Each subject underwent a pre-operative (P0) and post-operative (P1) gait analysis study before and 1 year after orthopedic surgery as part of the routine clinical decision making process, which included the following: physical examination, video analysis, motion analysis, force and EMG data collection[2]. Subjects were then recruited to participate in another gait analysis study at least 10 years (11±2) after surgery (P2). An additional two subjects (2 sides) were located through retrospective chart review who had pre and post-operative tests as well as a 10 year post-operative test as part of routine clinical care, for a total of 27 sides. Subjects were excluded if they had any intervening surgery between P1 and P2 or if they had a tibial derotation osteotomy. Kinematic, kinetic and clinical variables collected at P0, P1 and P2 were compared using a repeated measures analysis of variance (ANOVA) with Duncan's multiple range post hoc testing.

Results

Average age of subjects at P0 was 8±3, P1 was 9±3 and P2 was 20±4 years. Statistically significant improvements from P0 to P1 were seen in decreased mean passive hip internal rotation (75º±12 to 50º±10), increased mean passive hip external rotation (26º±15 to 41º±13) and decreased mean femoral anteversion (48º±12 to 25º±12), (Fig. 1). There was also a significant difference between P0 and P2 for passive hip internal rotation and

femoral anteversion but not passive hip external rotation. There were statistically significant kinematic changes seen from P0 to P1 and P0 to P2 in mean hip rotation in stance and mean foot progression in stance (Table 1). Walking velocity was not significantly different among the three analyses.

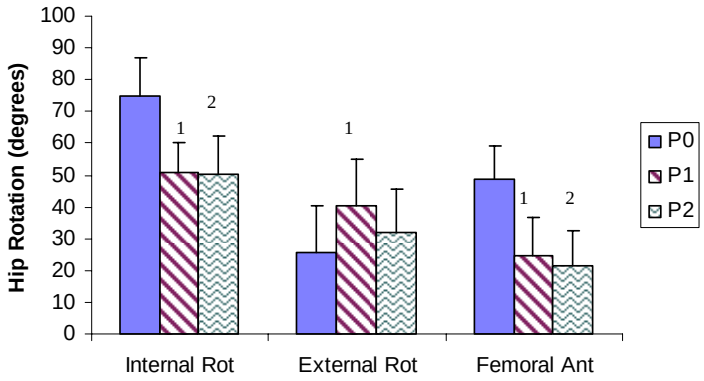

Fig. 1. Comparison between clinical examination measures of passive hip rotation and estimate of femoral anteversion among the three gait analyses. ([1] significant difference between P0 and P1, $p<0.05$; [2] significant difference at between P0 and P2, $p<0.05$; rot = rotation, ant = anteversion)

	Hip rotation stance (deg)	Foot progression stance (deg)	Walking velocity normalized to height
P0	20º±10	8º±18	0.7±0.2
P1	0º±9[1]	-14º±12[1]	0.7±0.2
P2	3º±16[2]	-16º±16[2]	0.6±0.2
typical	3º±1	-11º±8	0.9±0.1

Table 1: Comparison of selected kinematic variables for the three gait analyses. ([1] significant difference between P0 and P1, $p<0.05$; [2] significant difference at between P0 and P2, $p<0.05$; deg = degrees)

Discussion

Statistically significant changes in both clinical and gait variables demonstrate that femoral derotation osteotomies are effective in reducing excessive passive range of internal hip rotation, increasing limited passive external rotation and decreasing excessive femoral anteversion. These changes led to an improved gait pattern including decreased internal hip rotation and associated decreased internal foot progression during the stance phase of gait. All changes from P0 to P1 were maintained at P2, except for changes in passive hip external rotation, demonstrating lasting effects of this procedure over 10 years. While passive external hip rotation improved initially, it showed a trend toward decline in passive range of motion over time. There was no decline in functional capacity for ambulation over time as demonstrated by stable walking velocity. These findings indicate that persons with CP who experience internal hip rotation during gait can expect improvement in this pattern following FDO and can expect these improvements to be maintained in the long-term.

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COST COMPARISON OF INITIAL AND SUBSEQUENT SURGERIES BETWEEN PATIENTS WITH AND WITHOUT PRE-OPERATIVE GAIT ANALYSIS

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INTRODUCTION

One of the potential benefits of clinical gait analysis is that it may reduce the need for serial surgeries, thereby reducing long-term costs. Last year, we reported a reduction in the rate of additional surgery in ambulatory children with cerebral palsy (CP) who had undergone pre-operative gait analysis compared with peers who underwent surgery without gait analysis [1]. Patients who had undergone gait analysis had more procedures done during the index surgery, but fewer subsequent surgeries. In this study, we examine the cost implications of these findings.

CLINICAL SIGNIFICANCE

This study demonstrates that patients who undergo pre-operative gait analysis testing have a lower rate of subsequent surgery and similar surgical costs compared with peers who have not had gait analysis. Thus, patients with gait analysis experience less disruption of their lives due to surgery and rehabilitation without increasing costs. Documenting such benefits is important to justify the use of clinical gait analysis in a healthcare climate that emphasizes evidence-based practices and cost-benefit analysis.

METHODS

A retrospective review was conducted to identify all ambulatory patients with CP who had undergone lower extremity orthopaedic surgery at our hospital between 1991 and 2005 with at least a 6-month follow-up. Baseline demographic, clinical, and surgical data were abstracted, along with whether the patient had undergone a pre-operative gait analysis study. The patient characteristics examined included gender, age, race, CP subtype, ambulatory status, and GMFCS level. The primary outcomes included the number of surgical sessions and procedures for each subject and the direct costs associated with these surgeries.

Costs were calculated using a standardized protocol since the actual charges for a surgery could vary due to changes in medical and billing practices over time. The direct costs in our study included: 1) surgeon's fees based on Resource-Based Relative Value Scale (RBRVS) units with the standard 50% discount for all procedures after the first procedure, 2) anesthesia and operating room fees based on an estimated usage time of surgery duration plus 0.5 hour, 3) hospital stay fees based on a length of stay of 1 day for unilateral soft tissue, foot, or patella surgery, 3 days for femoral varus derotation osteotomy with concomitant procedures, and 2 days for all other surgeries, and 4) physical therapy fees based on 33 visits for soft tissue surgery and 65 visits for surgery involving bone procedures. All costs were calculated using our hospital's 2007 charge-master and standard billing practices.

Patient characteristics and outcomes were compared between patients who had undergone gait analysis testing before the index surgery (GA group) and patients who had surgery without gait analysis testing (NGA group). The general linear model (GLM) was used to compare costs between the two groups adjusting for significant risk factors.

RESULTS

The study cohort consisted of 440 ambulatory patients with CP; 308 had undergone gait analysis testing before their index surgery (GA group), and 132 had surgery without gait analysis (NGA group). Patients in the GA group were significantly older at the first surgical session, had lower (more functional) GMFCS levels, had their first surgery in later years, and had shorter follow-up (P<0.001). These factors were therefore adjusted for in the outcome assessments.

Costs for the initial surgery were higher in the GA group ($43,130 for GA vs. $39,818 for NGA, P<0.001) (Table 1). However, the costs associated with subsequent surgeries were higher in the NGA group ($3,248/person-year for NGA vs. $977/person-year for GA, P<0.001). There was no significant difference in the total cost per person-year after adjusting for covariates ($20,360/person-year for GA vs. $21,381/person-year for NGA, P = 0.58).

Table 1: Comparison of costs between GA and NGA groups

Cost Indicator	Statistic	GA Group (N = 308)	NGA Group (N = 132)	P-value
Cost in first surgical session	Adjusted Mean ± SEM	$43,130 ± 432	$39,818 ± 679	<0.001
	Median	$41,582	$39,872	
	Range	($26,001 - $66,464)	($26,001 – $71,415)	
Additional cost per person-year	Adjusted Mean ± SEM	$977 ± 304	$3,248 ± 478	<0.001
	Median	$0	$0	
	Range	($0 - $26,800)	($0 - $45,678)	
Total cost per person-year	Adjusted Mean ± SEM	$20,360 ± 859	$21,381 ± 1,351	0.577
	Median	$17,806	$11,967	
	Range	($3,298 - $89,619)	($2,229 - $94,638)	

* Means were adjusted for age, GMFCS level, and year of first surgery using parametric GLM and compared between groups using non-parametric GLM.

CONCLUSIONS

Patients with CP who underwent pre-operative gait analysis had more procedures done during the initial surgery [1] and therefore higher initial costs. However, annual costs after the initial surgery were reduced substantially due to a decrease in the rate of additional surgery. The overall result was no significant change in the total costs. Based on the results of Õunpuu et al. [2], which showed much higher costs for common surgical combinations when performed in a staged manner compared with single event multilevel surgery (SEMLS), we expected subjects with gait analysis to have lower overall costs. Nevertheless, without increasing costs, there was a benefit of gait analysis to the patients since the reduced rate of subsequent surgery means less disruption to patients' lives.

REFERENCES: [1] Wren et al. GCMAS, 2008, 126-127. [2] Õunpuu et al. GCMAS, 2008, 160-161.

The Effect of Static Standing Posture on Dynamic Walking Kinematics: Comparison of a Patella Marker Versus a Thigh Wand

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Introduction

3-D computerized gait analysis typically has used a thigh wand affixed to the lateral and distal part of the thigh [1] to measure hip rotation in walking. Kinematic hip rotation measurements using a thigh wand have been shown to have large variability [2]. Soft tissue artifact has been proposed as a primary source of that variability. Marker configurations using a distal thigh are able to account for only about 50% of the actual range of motion during a standing internal/external hip rotation task [3, 4] where a patella marker, in lieu of a thigh wand, was reported to detect 98% of the actual hip rotation range of motion [4]. It is not clear how different marker configurations impact hip rotation for gait. The patella marker and often the thigh wand are technical markers that require a data capture in static standing using a Knee Alignment Device to determine rotation values (offsets) to place these markers into the anatomical coordinate system. If a marker does not accurately represent the position of the hip during standing data capture, the technical marker will not be placed into the correct anatomical plane for the dynamic trial. This is particularly problematic if the static and dynamic positions of the hip vary from one another. The purpose of this study was two-fold: 1. to determine how much the static standing posture affects kinematic gait data of hip rotation and 2. to determine if a thigh wand or patella marker performed more consistently when subjects stand with a variety of hip rotation positions.

Statement of Clinical Significance

Hip rotation during gait is an important kinematic variable used in clinical decision making and particularly in planning femoral derotational osteotomies. Therefore, accurate hip rotation calculations and representations are critical.

Methods

Following local IRB approval, 10 typical adult subjects participated in this study (5 females, 5 males; mean age = 34 years). Static and kinematic data were collected using an 8-camera VICON 612 system. Static and dynamic processing was completed using Plug-In-Gait. For static trials, a typical Newington marker set was placed on the subject. Thigh markers were mounted to a base with a 5cm long wand and affixed to the distal thigh with 2-sided tape and wrapped with Coban for secure placement. An additional marker on a standard base without a wand was placed in the center of the patella. For static trials, subjects stood on an elevated platform weight bearing on the non-test side with the test side non weight bearing. Subjects were given a forearm crutch to closely simulate bilateral weight-bearing. Three static trials were then captured: 1. typical standing, self-selected hip rotation; then in random order, 2. hip internal rotation; 3. external hip rotation “as much as comfortably possible”. For all trials, the knee was in full extension. Each of the 3 static trials were processed twice, once using the thigh wand and a second time using the patella marker as opposed to the thigh wand. Thigh rotation offset values for each of the 6 conditions were recorded. The subject then walked typically for 3 trials with 1 of the trials selected randomly for analysis. The single gait trial was processed 6 times (applying 2 marker configuration offset values for each of the 3 static positions) recording mean stance phase hip rotation for each condition.

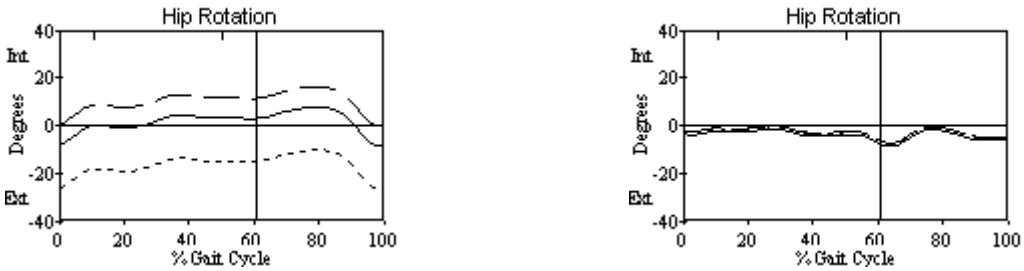
Results

The mean thigh rotation offsets calculated from the static trials showed significant differences with hip rotation position for the thigh wand whereas offsets did not changed significantly with the patella marker (Table 1). The mean dynamic hip rotation determined in stance phase was also significantly different for the thigh wand but not for the patella marker when the static hip posture varied (Table 1, Figure 1).

Table 1: Rotation offset-static trials and dynamic hip rotation values based on 3 static hip rotation postures and 2 marker configurations (patella vs. thigh wand).

	Rotation offset – static trial		Mean hip rotation – gait	
Static Posture	Thigh Wand	Patella Marker	Thigh Wand	Patella Marker
Typical Hip Rot	-9.1*	-93.9	-2.7* †	-7.1
Int Hip Rotation	-2.0*	-93.6	7.2* †	-6.8
Ext Hip Rotation	-19.5*	-93.7	-17.5* †	-6.9

*Significant difference between all 3 standing hip positions ($p < 0.0001$).
†Significant difference between patella and thigh wand for mean stance hip rotation in gait ($p < 0.05$).



(a) Thigh Wand **(b) Patella marker**

Figure 1. Example hip rotation, Subject 8: thigh vs patella markers. All traces are same gait trial and differ by static posture: typical (solid), int hip (large dash), ext hip (small dash).

Discussion

Static standing posture greatly affected the dynamic hip rotation kinematics when using a thigh wand. Errors presented because the distal thigh wand does not fully track the quantity of hip rotation. Dynamic hip rotation when utilizing a patella marker in lieu of a thigh wand was not effected by static hip posture. If a thigh wand is to be used in clinical practice, it is necessary that patients stand in a hip rotation posture that is equivalent to hip rotation position used in gait. This can be very difficult because it requires clinicians to have a priori knowledge of the gait hip rotation before testing. Also, patients may use different strategies in static standing than with walking posture. A patella marker detects nearly the full range of hip motion [3, 4], therefore the calculated rotation offsets to align technical into the anatomical coordinate system were consistent with varying static hip postures. This resulted in gait hip rotation that was largely unaffected by static posture with the patella marker. In conclusion, to determine gait hip rotation, it is critical that subjects stand with equivalent hip rotation in a static trial as in gait when a thigh wand is used. Alternatively, it is recommended that a marker on the patella be used instead of a thigh wand so standing static posture is not as critical and gait hip rotation is more consistently determined.

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GAIT PATTERNS IN CHILDREN WITH LIMB LENGTH DISCREPANCY

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Introduction

Very few articles describe the compensations in gait caused by limb length discrepancy (LLD) [1, 2, 3]. Song, et.al explored kinematic and kinetic variables utilizing work equalization as a marker of successful compensation for LLD. They found no difference in strategies based on the location of pathology. The purpose of this study was to define the various gait patterns in patients with LLD and the impact of these compensations on gait kinetics.

Statement of Clinical Significance

By defining the mechanical effects of LLD, treatment can be tailored to each patient to maximize joint and limb function, not just equalization of length.

Methods

Forty-three children (mean age 12.9 ± 3.7 yrs) with LLD greater than 2 cm were evaluated in the motion lab using a VICON motion system with two AMTI force plates. Etiologies included Legg-Calve-Perthes, developmental hip dysplasia, growth plate damage due to infection or trauma, congenital shortening of the femur or tibia , and syndromes creating shortening of the limb. Evaluation included physical exam and 3-D motion data generated using the model described by Vicon Clinical Manager (VCM). For data analysis, three representative trials were processed with the Plug-in Gait (PIG) lower body model using the "VCM spline" filter.

Walking strategies were identified by visual review. A kinematic threshold of two standard deviations away from normal values was used for inclusion in each group. Strategies included 1) pelvic obliquity with the short side lower (<-1.5 º) 2) flexion of the knee of the longer leg in stance (>5.2 º) 3) plantarflexion of the ankle on the shorter leg through the gait cycle (<0 º) , and 4) early plantarflexion crossover of the shorter limb (plantarflexion crossover occurred before 35% of the gait cycle). Variables were extracted into Excel using PECS (Vicon Motion Systems). The mean of the three trials was used for analysis.

Scanograms were used to establish lengths of the femur and the lower leg including the foot. The percentage difference for the subject (%LLD) was calculated as the leg-length between the two sides divided by the length of the long side.

The total mechanical work over the stride was the sum of the positive work and the absolute value of the negative work in all planes (1, 4). Paired t-tests were used to analyze the work differences between the short limb versus the long limb. Unpaired t-tests were used to compare between the different groups (short tibias, short femurs and controls).

Results

		Short side – Ankle Strategies		Long Limb	Both
Short Segment	**Pelvic Obliquity only**	**Equinus only**	**Vaulting only**	**Flexed knee Only**	**Multiple Strategies**
Femur (n=15)	6%	13%	27%	20%	33%
Tibia (n=9)	56%	11%			33%
Both (n=19)	26%	11%	16%		47%

Table 1: Strategy by location.

	Work at the Hip		**Work at the Knee**		**Work at the Ankle**		**Total Work**	
	Short	Long	Short	Long	Short	Long	Short	Long
Femur Short Only (n=19)	0.49	.49	0.44	.47	0.58	.49	1.52	1.44
Tibias Short Only (n=9)	0.57	.66	0.40	.56	0.28	.53	1.24	1.75
Both Short (n=15)	0.45	.60	0.36	.56	0.39	.55	1.21	1.71
Normal (n=10)	0.50		0.48		0.36		1.34	

Table 2: Work distribution

Distribution of single strategies for the group included: pelvis (11), equinis (5), vaulting (7), knee flexion (3); 17 subjects used multiple strategies. If the discrepancy was in the femur, patients chose a more distal compensation strategy, utilizing ankle movements which resulted in more work at the ankle joint on the short limb compared to normal ($p<.0001$). All subjects with tibia shortening showed pelvic obliquity (3 combined with knee flexion), which caused more work at the hip joint on the short limb compared to normal ($p<.01$). Total mechanical work on the uninvolved limb was above normal for all groups ($p<.0001$) (Tables 1 & 2).

Discussion

Our study contradicts previous literature that found no difference in strategy based on location of the shortening and also, a higher number of children with pelvic obliquity than previously described [1]. It appears different compensation schemes are employed by patients with LLD. The increase in work may have long term implications for management. Future studies will include changes in kinematics and work, after intervention. Better understanding of postoperative changes from different surgical methods may provide more insight for preoperative planning and may lead to a more satisfactory outcome for specific patients.

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Acknowledgements
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GAIT ASYMMETRY FOLLOWING THA - ANTERIOR VS LATERAL APPROACH

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INTRODUCTION

Several surgical approaches to total hip arthroplasty (THA) are available for patients, though questions remain regarding which surgical technique produces the greatest results. Due to pain and weakness in the involved hip, patients undergoing THA often demonstrate a pattern of prominent limping or asymmetries in their gait (Cichy et al., 2008). The purpose of this study was to examine which surgical approach, lateral or anterior, minimized short term limping in THA patients. It was hypothesized that prior to THA, both the lateral and anterior groups would display asymmetric gait patterns when compared with the control group. It was also hypothesized that compared to the lateral group, the anterior group would display less asymmetry in gait pattern both six weeks and 16 weeks post-op.

CLINICAL SIGNIFICANCE

The incidence of post operative limping following THA has been shown to differ based on surgical approach (Masonis and Bourne, 2002). With the increase in THA surgeries and hospitalization charges, the quantification of gait asymmetries can provide patients and surgeons with information to better understand postoperative hip function.

METHODS

Twenty three patients, including 12 patients undergoing anterior THA (mean age = 56.9 ± 3.4 years, BMI = 32.0 ± 5.1 kg/m^2, 7 males) and 11 patients undergoing lateral THA (mean age = 57.0 ± 7.3 years, BMI = 31.1 ± 4.1 years, 9 males) were recruited for this study. All patients had unilateral osteoarthritis of the hip, with no prior lower limb surgery, neurological or musculoskeletal deficits. Patients were tested during three time periods: pre-surgery, 6 weeks post-surgery and 16 weeks post-surgery. Ten control subjects (mean age = 59.9 ± 5.3 years, 26.3 ± 3.9 years, 5 males) with no lower limb joint surgery, arthritis or vertigo were also recruited from the community. Control subjects were tested during two visits one month apart.

Subjects were asked to walk at a self-selected speed across a 10 meter walkway during all visits. Marker trajectories of 29 markers placed on bony landmarks (Chou et al., 2003) were captured at 60 Hz using an 8-camera motion analysis system (Motion Analysis Corp., Santa Rosa, CA). Single limb support time, step length, and pelvic obliquity were calculated throughout the gait cycle. The symmetry index for support time and step length was determined as previously reported (Zifchock et al., 2008). A two-way ANOVA with repeated measures was used to detect differences in group and visit on symmetry indices and pelvic obliquity. Statistical analysis was performed using SPSS 14.0 (SPSS Inc., Chicago, IL).

RESULTS

A group* time interaction was found for single limb support time, with anterior and lateral THA groups demonstrating greater symmetry between limbs by 16 weeks post-surgery, when compared to pre-surgery (Table 1; $p < .001$ and $p = .018$, respectively). By 16 weeks post-surgery, the anterior group had reached the level of controls ($p = .702$), while the lateral

group remained more asymmetric than controls (p = .003). While no group*time interaction was found for step length, across all times the lateral group was significantly different than controls (p = 0.003).

Across all visits, at the moment of weight acceptance of the surgical limb, the anterior and lateral groups showed a significantly smaller pelvic obliquity than controls (Fig 1; p = 0.047 and p = 0.012, respectively). No such group differences were found for pelvic obliquity at weight acceptance of the non-surgical limb (p = 0.085). Neither the anterior or lateral group displayed improvement from pre-surgery to 16 weeks post-surgery for pelvic obliquity (p = 0.525).

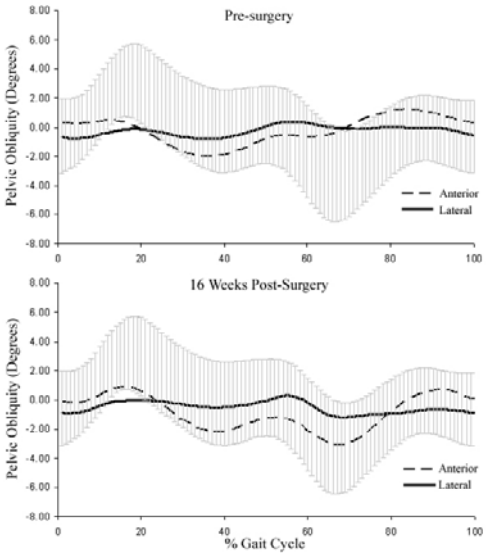

Fig 1: Pelvic obliquity across 100% of the surgical limbs gait cycle for THA compared to controls

Table 1: Symmetry index (SD) between limbs

	Control [a]	Anterior THA	Lateral THA
Single Limb Support			
Pre-surgery	2.3 (1.6)	11.9 (9.5) *	12.3 (11.9) *
16 weeks post		2.0 (1.6) †	5.6 (3.3) * †
Step Length			
Pre-surgery	2.7 (2.2)	10.5 (12.2)	9.9 (8.4)*
16 weeks post		4.6 (2.7)	8.2 (7.8)*

[a] Control values were averaged across the two visits since no differences were found across time.
* Significant difference between patient group and controls (p < .05)
† Significant improvement from pre-surgery (p < .05)

DISCUSSION/CONCLUSIONS

In the past, step length, single limb support time pelvic obliquity have all been used to determine gait asymmetry (McCrory J. et al, 2001; Cichy et al., 2008). With these measures, the present study showed that prior to THA, both surgical groups displayed asymmetric gait patterns. These patterns may be due to adaptive compensatory behaviors and pain during ambulation. While the anterior group did demonstrated greater improvement by 6 weeks post-surgery, by 16 weeks post-surgery though, both THA groups approached control levels, with reduced asymmetries between their surgical and uninvolved limbs. While most clinical studies determine limp through a subjective Trendelenburg test, asymmetries can be found objectively using the gait measures presented in this study.

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Functional Assessment of Stair Negotiation Following Total and Resurfacing Hip Arthroplasty

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INTRODUCTION:

Total hip arthroplasty (THA) and resurfacing hip arthroplasty (RHA) are indicated treatments for patients who suffer from advanced osteoarthritis (OA) of the hip joint. Common daily activities such as obstacle clearance and stair climbing must be recovered following surgery for effective community ambulation. Though gait assessment is often used, climbing and descending stairs is considered a more sensitive tool for measuring functional ability post-operatively. Stair negotiation also identifies the importance of intra- and interlimb relationships, which can be used to compensate for pathologies. Since step length is limited during stair negotiation, greater focus can be placed on kinetic and temporal variables. Studies have used 'step up' models where a subject with OA is asked to climb onto an instrumented box and only limited studies have used a full stair case (McFayden 2001). Data on stair negotiation following RHA or THA is currently unavailable in published literature. Comprehension of functional recovery of common activities following HA is needed for improved prognosis, prescription of rehabilitation post-operatively, and future understanding of in situ function of resurfacing and total joint arthroplasty. This study characterizes stair ability pre-operatively and 3mo following RHA and THA.

METHODS:

This study was performed at the SHRI-CORE Motion Analysis Lab in Sun City West, AZ. The study received IRB approval from the Sun Health Institutional Review Board and all subjects signed informed consent and HIPPA forms. Subjects were recruited from physician's clinics at The CORE Institute. Subjects with advanced OA of the hip scheduled to receive surgical intervention were recruited for this study.

Participation in the gait lab was requested pre-operatively and at 3 months post-operatively. Assessment was performed using a 10 camera passive marker motion analysis system (Eagle-4, Motion Analysis) and an instrumented 4 step stair case (AMTI). Three dimensional kinetic and kinematic data was collected. A skeletal model was created using a modified Helen Hayes marker system. Subjects were instructed to ascend (SA) and descend (SD) stairs at a self selected speed, in a foot-over-foot manner. A hand rail was available for safety, though not promoted for use. Demographics of study participants are listed in Table 1.

Kinetic and kinematic data was post processed using EVaRT 5.0.1 and OrthoTrak 6.23 biomechanical software. Matlab 8.0 was used for custom plotting. The SPSS statistical package was used to determine correlations between survey and quantitative results, and run one-way ANOVA and t-tests for intra-subject and group comparisons at all study time points. Homogeneity of variances was determined and post hoc testing was performed if needed.

Table 1: Subject Demographics

	RHA n=11	**THA n=13**	**CONT n=12**
Age	48.7 ± 4.8	58.5 ± 12.1	49 ± 6.7
Sex (M/F)	9M/2F	8M/5F	5M/7F
Height (cm)	172 ± 9.7	168.7 ± 10.2	166.1 ± 11.3
Weight (kg)	82.5 ± 17.7	82.2 ± 18.4	69.5 ± 15.3

RESULTS:

Significance differences were noted between both patient groups in comparison to controls pre-operatively: cadence (p<.001 SA and SD); hip extension (p=.004 SA and SD); vertical ground reaction force (GRF) (p=.011 SA, p<.001 SD); step width (p=.039 SA, p=.009 SD); single support time (p=.028 SA, p=.001 SD). At three months, both groups indicated improvements with similar values to controls. Higher peak flexion and abduction moments and improved hip extension was noted in the RHA group in comparison to the THA group. The RHA group maintained a slower velocity and had significantly (p=.04) less hip power than the control and THA groups. Both groups maintained lower vertical GRFs than controls during SA and SD.

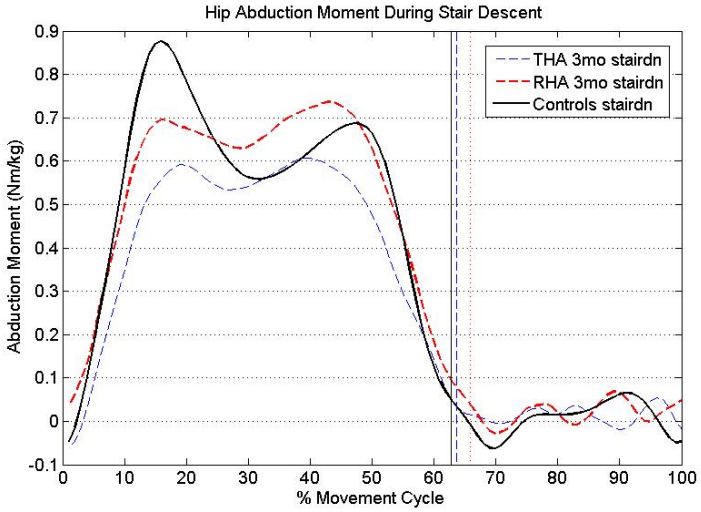

Figure 1. Frontal plane hip kinetic result during stair descent.

DISCUSSION:

Stair negotiation was improved following surgical intervention in both RHA and THA groups. This may be due to reduced pain and increased range of motion during joint loading. Greater forces were placed in the hip during SA. During SA the RHA group presented greater hip extensor ability but significantly less power. Both groups had longer double support times than controls for the injured limb. Step width was corrected in the THA group at 3 months but remained for SD in the RHA group. Peak hip abduction angles and moments during all stair trials indicated a larger improvement in the RHA group. Moments in both groups indicated premature initiation of forward continuance of the foot and a shortened period of weight acceptance. This may be indicative of remaining pain and reduced confidence in the limb. Stability of the upper extremities and trunk and balance of the center of mass is crucial to completion of stair tasks. Both types of artificial joints require removal of the acetabular labrum and invasion of surrounding musculature. Rate of recovery of proprioception and muscle mechanics may help explain differences seen between patient groups. Different strategies were seen between THA and RHA groups to complete the task similar to controls, though neither group reached full recovery which was indicated by remaining deficits in other planes of movement. Studies with longer duration are needed to determine full recovery of stair function following RHA and THA.

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ACKNOWLEGMENTS: support for this study was provided by Smith & Nephew.

The role of un-opposed Quadriceps Forces in Loading the Anterior Cruciate Ligament during Jump Landing

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INTRODUCTION

It has been suggested that application of an unopposed and large quadriceps force may result in excessive loading of the anterior cruciate ligament (ACL) and its subsequent rupture. This so-called quadriceps-pull mechanism has received significant attention as a possible sagittal plane mechanism of non-contact ACL injury[1]. Theoretically, under the application of the quadriceps force, the patellar tendon pulls the tibia anteriorly while the patella pushes the femur posteriorly and if this action is unopposed (no hamstring force) it could result in ACL injury. This proposed mechanism has been controversial, as other researchers endorse a coronal plane mechanism of ACL injury (dynamic valgus), asserting that the quadriceps pull mechanism is not viable for two reasons: 1) the suggested critical quadriceps force is too high and unachievable at low knee flexion angles and 2) as the quadriceps force increases, the tibia moves anteriorly but the anterior component of the patellar tendon force decreases as the tendon becomes more vertical[2]. At this point, no further anterior tibial translation can be achieved by the tendon. In response, some suggest that if the un-opposed quadriceps pull is combined with a dynamic ground reaction force (for instance during jump landing), the combination may injure the ACL. To address this controversy, in-vitro simulators that model specific athletic activities (such as jump landing) and integrate muscle forces may be most effective. In this study, one such simulator was used to investigate the role of an unopposed quadriceps force on ACL strain during the landing phase of a jump. We hypothesize that as quadriceps pre-activation force (QPF) increases in anticipation of landing, the pre-landing strain in the ACL increases but the strain during the actual landing phase (landing strain) decreases. We further hypothesize that the total strain in the ACL (pre-landing strain + landing strain) remains constant and is independent of the quadriceps load.

METHODS

A jump landing simulator was designed for dynamic loading of cadaveric knees with ability to apply dynamic muscle forces using servo-electric actuators. The system allows for the positioning of the desired anterior-posterior hip position, hip flexion angle, and knee flexion angle. A linear actuator attached to the patella using cables and steel wires simulates the action of the quadriceps muscle. A differential variable reluctance transducer (DVRT) was mounted on the ACL to monitor strain during quadriceps pre-activation as well as during impact. An impact-lever apparatus is utilized to impulsively load the tibia (upward), simulating a landing ground reaction force.

Nine frozen cadaveric knees were used in this study. Each knee was thawed and examined for joint disease and damage to the ligaments and the knee capsule. Each knee was placed in the simulator at a flexion angle of 20°. The ACL was instrumented with a DVRT at that position (the zero strain reference position). The QPF was increased incrementally after

each test using the electric actuator attached to the patella via a steel cable that was aligned in the direction of the femur. The range of applied QPF was from 25 N to 700 N. In two knees the QPF range was extended to 1500 N. When the desired QPF level was reached, the strain in the ACL (pre-activation strain) was measured. At this point the knee was loaded impulsively upward at the simulated ankle joint using the impact lever system. The strain in the ACL was also measured during the landing phase. Correlation analysis was performed for individual knees as well as the averaged responses of all knees to assess the extent and nature of relationships between i) QPF vs. ACL pre-landing strain, ii) QPF vs. ACL landing strain, and iii) QPF vs. total strain (pre-landing + landing strain). The statistical significance of each correlation was assessed.

RESULTS

None of the ACLs were injured after simulated jump-landing. In each individual knee, there was a strong and positive correlation between QPF and ACL pre-landing strain. There was also a strong but negative correlation between QPF and landing strain in seven out of nine knees (the other two knees showed weak correlations). There was no correlation between QPF and total ACL strain. When the responses were averaged for all knees, the R^2 value for the QPF vs. total ACL strain was low ($R^2 = 0.06$), indicating that QPF had little effect on the variation in total ACL strain . The ensemble averaged total ACL strain was approximately 5% for all values of QPF during the simulated jump landing process.

DISCUSSION

Pre-activation of the quadriceps in anticipation of landing increases the ACL pre-landing strain but it decreases the landing strain. Decreased landing strain as QPF increases is indicative of a more stable joint. The total ACL strain remains constant throughout the range of QPF. This result was observed in eight out of nine individual knees and in the averaged behavior of all knees. This suggests that each individual knee may have a “signature” strain for the jump landing activity. As total strain remains constant with increased QPF, it is apparent that there is no mechanism by which quadriceps could cause ACL injury during landing from a jump, even when un-opposed by other muscle forces and when aided by the ground reaction force. Although the total strain in the ACL remains constant, low QPF is more dangerous from an injury perspective. This is because under low QPF the landing strain is high. Higher ACL landing strains represent a less stable knee during the dynamic process of landing. If other factors such as inadequate response of the hip [3] are coupled with low QPF, the ACL may experience excessive strain and even failure.

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A Comparison by Motion Analysis of Anterior Cruciate Ligament Reconstruction using a Patellar Tendon versus a Hamstring Tendon Autograft

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INTRODUCTION

Two of the more common procedures used to reconstruct the injured or torn anterior cruciate ligament (ACL) of the knee involve the use of an autogenous graft from either the patellar or hamstring tendons [1]. Previous data reported in the scientific literature indicate that knee biomechanics during gait differ between graft type [2]. However, there's a lack of quantified data that compare the effects of these two procedures on lower extremity function during high-impact activities such as running, cutting, and landing from a jump, which are typically associated with ACL injuries. Hence, the purpose of this study was to compare lower extremity function through motion analysis of these sport-specific maneuvers following ACL reconstruction with a patellar graft versus a hamstring graft.

CLINICAL SIGNIFICANCE

With an estimated 80,000 ACL injuries occurring annually [3], the capacity to identify the mechanical differences between two of the more common ACL reconstruction techniques has extensive clinical implications.

METHODS

A total of 13 subjects with unilateral ACL tears participated in this study after providing written informed consents. The mean age, height, and weight of all the participants were 19 ± 7 years, 170 ± 15 cm, and 73 ± 20 kg, respectively. Nine participants ultimately underwent an ACL reconstruction with a bone-to-bone patellar tendon (BTB) autogenous graft while four participants underwent reconstruction with a hamstring (STG) autogenous graft. All participants postoperatively underwent the same rehabilitation protocol. Each participant completed the International Knee Documentation Committee (IKDC) form to document relevant symptoms, signs, and function at each analysis. Each participant underwent the same biomechanical testing protocol during each of three visits: pre-operatively, at 6 months post-op, and at 12 months post-op. Each participant performed 5 "usable" running trials and 5 "usable" sidestep cutting trials while 8 visible-red cameras (Motion Analysis Corporation, Santa Rosa, CA) captured the joint kinematics defined by reflective markers placed strategically on the lower body. A "usable" trial was defined as one in which the subject's entire foot on the involved side cleanly hit a 600 x 400 cm^2 force platform to measure ground reaction forces and joint kinetics (torques). In addition, each subject performed two landing maneuvers, one in which the subject jumped off and landed back on the platform with the involved leg and another in which both legs were used. Knee joint moment data were normalized to percentage of body weight and height to allow for between subject comparisons and are expressed as external moments. A modified repeated measures analysis of variance (RMANOVA) was used to test differences in dependent variables postoperatively between the BTB and STG groups at a significance level of 0.05.

RESULTS

IKDC scores indicated that both groups exhibited similar improvements in pain reduction and subjective function. No significant differences were found in any biomechanical parameter between 6-month and 12-month visits in either group. However, both groups exhibited significant increases in knee rotational moments during cutting post-operatively (Figure 1). The BTB group displayed a significantly higher knee varus moment during double landing maneuvers than the STG group (Figure 2).

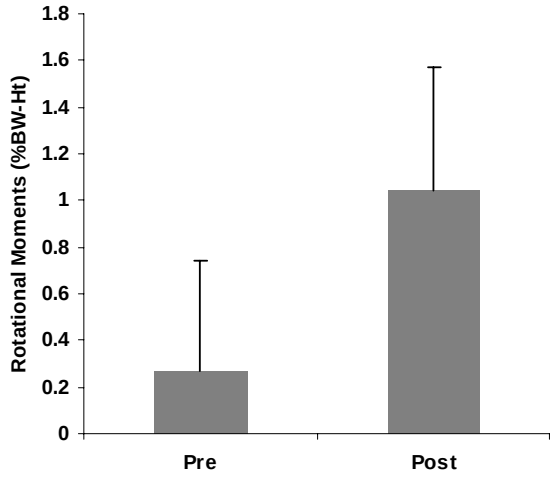

Figure 1: Knee rotational moments during cutting were significantly higher post-operatively (p=0.02).

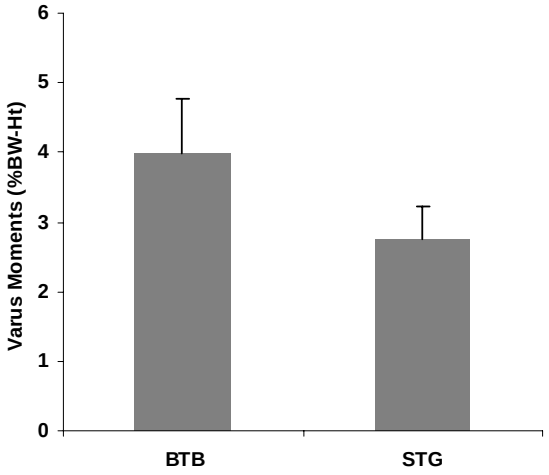

Figure 2: Knee varus moments during double landing were significantly higher in the BTB group (p=0.04).

DISCUSSION

Current literature suggests that there remain questions as to which autograft, patellar or medial hamstring, is more effective in restoring function lost to an ACL injury [4]. Although both reconstruction techniques yielded many similar biomechanical outcomes, the results of this study showed higher varus moments in patients with BTB grafts. Hence, donor site morbidity resulting from graft harvestation may compromise frontal plane stability during high-impact activities such as landing from a jump.

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PREHAB EFFECTS ON SIT-TO-STAND IN KNEE OA PATIENTS PRIOR TO TKA

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Introduction

The decision to undergo total knee arthroplasty (TKA) is common among knee osteoarthritis (OA) patients. TKA removes the painful joint; however, it inflicts tissue trauma that causes other, generally transient, inflammation and pain, which initially lead to decreased functional capacity. Rehab then seeks to improve capacity to hopefully greater than pre-surgery levels.

Applied to TKA, the concept of prehab[1] (similar to rehab, but prior to surgery) anticipates that a proper pre-TKA exercise regimen will increase pre-surgery functional capacity. The concept further anticipates that, after post-TKA functional declines, prehabbers will begin rehab with greater functional capacity, and will achieve greater post-rehab functional capacity.

Post-TKA functional assessment often focuses on overall task completion measures.[2] Other metrics (e.g. strength measures) are often not specific to any functional task. Measures that quantify the way that individuals perform functional tasks are less often considered in clinical post-TKA assessment of function. The ways in which specific tasks are performed, however, could have implications regarding development of future musculoskeletal pathology.

In the work reported here the investigators evaluated the prehab concept's initial portion (i.e. pre-surgery improvement via exercise), with respect to sit-to-stand function, among knee OA patients who had scheduled TKAs. The purpose was to evaluate whether a prehab exercise program could improve pre-surgery sit-to-stand function among knee OA patients who were scheduled for TKA. This evaluation not only utilized traditional clinical functional measures, but also involved biomechanical measures that quantified how sit-to-stand was performed.

Statement of Clinical Significance

Despite potential prehab related improvements in clinical functional measures, continued asymmetry in both control and prehab subjects' vertical ground reaction forces indicated that prehab did not affect the presence of knee OA patients' compensatory mechanisms during sit-to-stand. Persistence of compensatory mechanisms that tend to favor a less affected limb could be a factor in further degeneration of musculoskeletal tissue.

Methods

The present study sample consisted of 39 subjects who had decided to schedule unilateral TKA. Nineteen subjects (6 male, 13 female; mean age, height, and weight of 65±7 years, 166±10 cm, and 91±17 kg) were randomly assigned to a prehab group. The remaining 20 subjects (7 male, 13 female; 65±7 years, 169±11 cm, and 88±22 kg) were assigned to a control group (i.e. no prehab). All subjects signed an IRB approved consent form.

Subjects completed 2 identical data collection sessions that occurred ≥ 4 weeks prior to TKA, and ≈ 1 week prior to TKA. For each session's 1st sit-to-stand trial, each subject performed a

single rep with the unaffected and affected limbs on and beside a force platform, respectively. For the 2nd trial foot placements were reversed, and subjects were told to perform as many reps in 30 seconds as they were able. Subjects were also instructed to perform the 1st trial at the same pace as the 2nd. The 2nd trial rep count (n_{rep}) and peak, isokinetic, knee extensor torque ($\tau_{ext,peak}$), in %BW, were recorded as standard clinical functional measures. In addition, affected and unaffected limb, peak vertical ground reaction forces (VGRFs) were obtained from each trial's 1st rep. Unaffected limb minus affected limb VGRF ($VGRF_{unaff-aff}$) was computed to quantify symmetry (or lack of) in the way that sit-to-stand tasks were performed.

Results

Prehab group n_{rep} increases (session 1 to 2) were significantly greater than control group n_{rep} increases (2.2 vs 0.6; p=.03; Fig1). Prehab group's increases in affected side $\tau_{ext,peak}$ were not significantly greater than control group's (3.6 vs -5.4; p=.08; Fig1), although the designated α=.05 was approached. Prehab group $VGRF_{unaff-aff}$ changes were not significantly different from control group $VGRF_{unaff-aff}$ changes (.003 vs .034; p=.18; Fig1). For both groups, however, $VGRF_{unaff-aff}$ was significantly different from zero (p<.001; Fig 2) at each session.

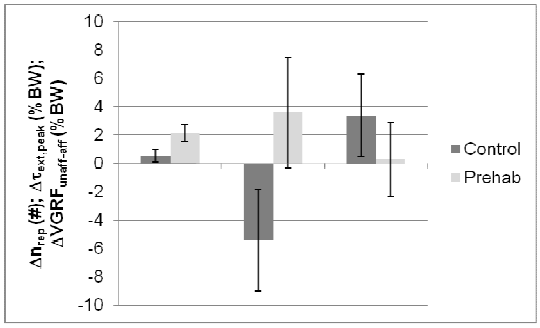

Fig 1. Session 1 to session 2 changes.

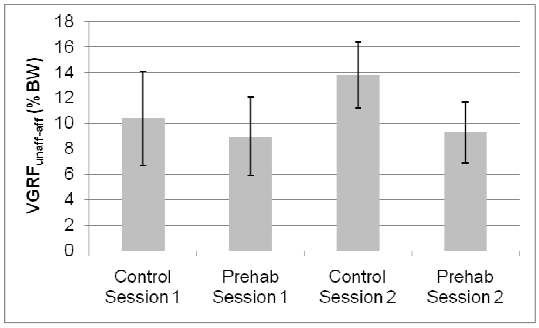

Fig. 2. Asymmetry by group and session.

Discussion:

Results suggest that prehab regimen can increase functional task completion measures (e.g. sit-to-stand reps in 30 sec.) and may increase strength measures (e.g. knee extensor torque) among knee OA patients prior to TKA. However, prehab did not affect the asymmetrical manner, as quantified by peak VGRFs, in which knee OA patients perform sit-to-stand tasks.

Compensatory mechanisms, particularly those associated with greater loading on one limb, likely develop to mitigate pain. Since affected side pain is not yet relieved prior to TKA, it would appear that the current prehab regimen of exercises (directed towards strength, flexibility and balance development) is unlikely to affect asymmetric loading compensations. Persistence of these compensations could result in accelerated deterioration of a "good" knee, and could be a factor in many knee OA patients undergoing a subsequent second TKA. Prehab, as well as post-TKA rehab, should consider including components directed towards modifying compensatory mechanisms.

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EFFECTS OF GAIT TRAINING WITH REAL-TIME BIOFEEDBACK ON KNEE ADDUCTION MOMENTS DURING WALKING IN ADULTS WITH KNEE OA

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Introduction

The development and progression of knee OA has been attributed to different factors, including abnormal biomechanical stressors (1, 4). Deviations in joint biomechanics during walking have been associated with progression of the disease, in particular, abnormally high frontal plane moments (1). Numerous approaches have been attempted in an effort to modify knee frontal plane moments with mixed results (3, 5, 6). The ability to apply patient-specific interventions has been advocated and offers the possibility of identifying and modifying the underlying pathomechanics. The purpose of this study was to investigate the effect of patient-specific gait training, with real-time biofeedback, on frontal plane knee moments during walking in adults with symptomatic knee OA.

Clinical Significance

The assessment of the effect of gait training with biofeedback and a muscle activation program will facilitate the development of effective rehabilitation interventions in adults with knee OA.

Methods

Fifteen subjects with radiographic KL grade ≥ 2 and ipsilateral knee pain or stiffness on most days of the month, participated in this study. During the first visit, all subjects underwent a physical and over-ground gait evaluation. The physical evaluation assessed strength, flexibility, and range of motion in each subject's trunk, legs, and feet. The gait evaluation was conducted along an 8 m walkway using a three-dimensional motion analysis system (Optotrak, NDI; Kistler) with subjects walking at 1.12 m/s. Gait data were processed using Visual 3D software (C-Motion).The gait information in combination with the physical evaluation information was used to establish patient-specific goals during gait training; with the possible addition of a home exercise program appropriate for correction of impaired flexibility and strength. Subjects participated in supervised treadmill gait training twice a week for 12 weeks. Subjects walked on an instrumented treadmill (Gaitway, Kistler) for 8 minute intervals with 3 to 5 minute rest periods. During the gait training, the subjects were provided with real-time biofeedback (Visual 3D) for correction of kinematic patterns that were thought to contribute to abnormal kinetics. Over ground kinetics and kinematics during walking were reassessed during the 8th, 16th, and 24th visits. The performance of exercise was checked periodically to encourage compliance.

Results

Fifteen subjects (9 female, 6 male; mean age 74.4 +/- 6.9; mean weight 81.09 +/- 14.11) took part in this study. The peak external knee adduction moment during gait was reduced by

0.29% from baseline (mean -.665 +/-.22) to third month of training (mean -.497 +/- .267). This reduction was significant after the first month of training (p = .0034) as shown on figure 1(V1 vs. V2). The first adduction moment peak was also reduced after the second month (p = .035). There was a significant overall reduction of the adduction moment of involved knee over the three months of training (p = .000104).

Discussion

The results of this study indicate that knee frontal plane moments could be reduced by supervised gait training with real-time biofeedback and specific and pre-identified exercises (2, 5). The feedback provided addressed abnormal kinematic patterns. The subject-specific goals were based on baseline information and monthly reassessments. The gait training mostly included modifications to the frontal, transverse, and/or sagital planes of trunk and pelvis movement at subjects' self selected walking speeds. Feedback was also provided during rest periods to reinforce corrected patterns. This data suggests participants' ability to dynamically modify trunk and pelvis kinematics to positively influence knee frontal plane moments.

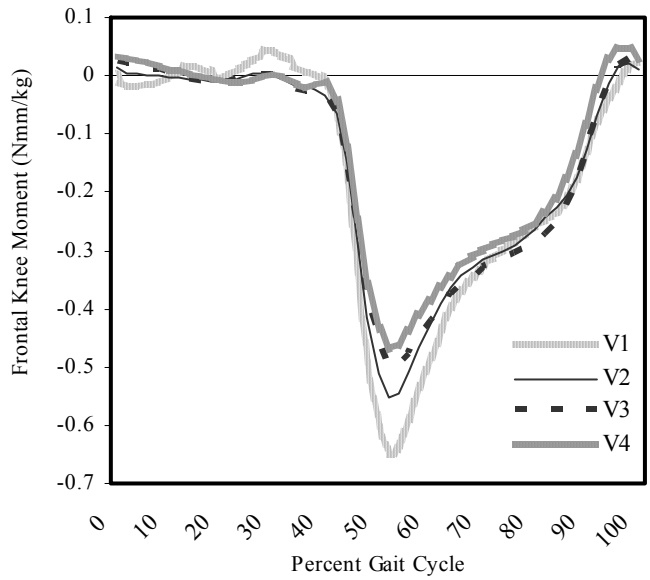

Fig.1: Reduction on the peak adduction moment during baseline (V1), and after 1st (V2), 2nd (V3), and 3rd (V4) month of gait training with real-time feedback.

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Acknowledgements

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Time	Thursday, March 12th, 2009
7:00 –8:00 am	Breakfast –*Exhibitor Area* (Registration 6:30am –3:00pm)
7:00 –8:00 am	Sunrise #2 –*Room 4* "Surgical Correction of Crouch Gait via Extension Osteotomy and Patellar Tendon Advancement" Tom Novacheck
8:15 –9:15 am	Keynote Address #2 –*Rooms 4–7* "Dynamic Walking: How to Locomote with Economy, Speed, and Stability" Arthur Kuo, PhD, University of Michigan 1 CME/CEU Credit
9:15 –10:25 am	Podium Presentations –*Rooms 4–7* *Session 4: Work and Energy* Moderators: Adam Shortland & Gordon Alderink 1 CME/CEU Credit

- External Mechanical Limb Work in Children with Cerebral Palsy
  M. J. Kurz, W. Stuberg, S. DeJong, and R. Harbourne
- Interrelationships Among Work, Energy Transfer, and the Cost of Walking in Children
  B. R. Umberger, S. Augsburger, J. Resig, D. Oeffinger, R. Shapiro, C. Tylkowski
- Kinematics, Metabolic Cost and External Work Done During Walking with a Forward Assistive Force
  B. Bennett
- Effect Of Walking Speed on Mechanical Recovery and Internal Work after Stroke
  B. Roewer, C. Henderson, D. Reisman, J. Higginson
- Muscle Contributions to Propulsion and Support During Running
  S. R. Hamner, C. T. John, J. S. Higginson, S. L. Delp
- Gait Classification Depends on Walking Speed
  M. M. van der Krogt, C. A. M. Doorenbosch, J. G. Becher, J. Harlaar
- Crouch Gait Patterns Derived from Cluster Analysis are Related to Clinical Parameters and Surgical Interventions
  A. Rozumalski, MS; M. H. Schwartz, PhD

10:25 –11:05 am	Break: Snacks in Exhibitor Area
11:05 am –12:15 pm	Podium Presentations –*Rooms 4–7* *Session 5: Simulation* Moderators: Mark McMulkin & Caroline Stewart 1 CME/CEU Credit

- Influence of Gait Analysis on Decision Making for Hamstring Lengthening Surgery
  T. A. L. Wren, R. E. Bowen, N. Y. Otsuka, A. A. Scaduto, L. S. Chan, M. Sheng, R. Hara, R. M. Kay
- Predicting Changes in Knee Flexion after Hamstrings Lengthening Surgery
  J. L. Hicks, M. H. Schwartz, S. L. Delp
- Accuracy of Six DOF Joint Kinematics During Normal and Pathological Gait: A Simulation Study
  A. Schmitz, D. Thelen, D. Bruening, F. Buczek
- Accelerations Produced by Gastrocnemius and Soleus During Crouch Gait
  K. Steele, A. Seth, J. Hicks, M. H. Schwartz, S. Delp
- Understanding Stiff-Knee Gait - A Dynamic Walking Approach
  M. M. van der Krogt, D. J. J. Bregman, S. H. Collins, C. A. M. Doorenbosch, J. Harlaar, M. Wisse
- Electrical Stimulation of Rectus Femoris During Pre-Swing Diminishes Knee Flexion During Swing
  A. Hernández, D. Thelen
- Mechanism of Improved Knee Flexion after Rectus Femoris Transfer Surgery
  M. D. Fox, J. A. Reinbolt, S. Õunpuu, S. L. Delp

12:15 –1:15 pm	Business Lunch –*Room 4*

1:15 –2:45 pm	Podium Presentations –*Rooms 4–7* *Session 6: Foot and Ankle* Moderators: Steve Piazza & Tom Novacheck 1 CME/CEU Credit

- Long Term Follow-Up after Comprehensive Clubfoot Surgery
  A. Graf; S. Hassani, MS; A. Flanagan; A. Caudill; J. Krzak; J. Long, PhD; K. N. Kuo, MD; G. F. Harris PhD; P. Smith, MD
- A Comparison of Children with a History of Clubfoot Which Underwent Surgical or Ponseti Treatment
  C. Church, J. Coplan, A. Thabet, D. Kowtharapu, N. Lennon, S. Marchesi, J. Henley, R. Starr, D. Mason, M. Belthur, J. Herzenberg, F. Miller
- Ankle Kinematics and Kinetics Following Non Operative Treatment for Clubfoot
  K. A. Jeans, MS; L. A. Karol, MD; R. E. Hawary, MD
- Plantar Pressures Vary Across Elliptical Trainers and Compared to Walking
  J. M. Burnfield, B. R. Barber, T. W. Buster, A. P. Taylor
- Correlation Between Plantar Pressure and Oxford Foot Model Kinematics in Clubfoot
  J. McCahill, J. Stebbins, C. Giacomozzi, T. Theologis
- Kinematic Analysis of Foot Motion in Children with Diplegic and Hemiplegic Patterns of Cerebral Palsy
  N. Lennon MS, PT; C. Church MPT; D. Taylor DPT; S. Marchesi PhD; J. Henley PhD; F. Miller MD
- Kinematics of the Unaffected Foot in Hemiplegic Cerebral Palsy
  J. Bates, MSc; J. Stebbins, Dphil; T. Theologis FRCS
- Comparative Motion of the Foot During Walking for Men, Women and Children
  J. Stebbins, T. Theologis
- A Comparison of Multi-Segment Foot Motion During Overground and Treadmill Walking
  K. Tulchin, MS; M. Orendurff, MS

2:45 –7:00 pm	Tour Option #1: Denver Art Museum and Wine Testing (same location) (3:00 -7:00 pm Exhibitor Tear Down - Rooms 1-3)
2:45 –7:45 pm	Tour Option #2: Coors Brewery and Red Rocks Amphitheater
2:45 –6:00 pm	Tour Option #3: Center for Gait and Movement Analysis (CGMA) – The Children's Hospital

External Mechanical Limb Work in Children with Cerebral Palsy

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Introduction

Several classic gait experiments have demonstrated that the lower limb behaves much like an inverted pendulum during the stance phase [1]. The inverted pendulum mechanics allow for the locomotive system to conserve mechanical energy through the exchange of potential and kinetic energy. This suggests that less work needs to be performed on the center of mass as long as the limb acts like a rigid strut. However, recent computer simulations indicate that a significant amount of muscular work is necessary to lift the center of mass during single limb support [2]. Furthermore, additional limb work is necessary during double support to redirect the center of mass from one pendular arc to the next [3]. The external work performed during double support has recently received considerable attention because it requires a coordinated effort by the trailing and leading limbs to redirect the center of mass. For this redirection to properly occur, the trailing limb must produce a sufficient amount of positive external work to balance the amount of negative external work produced by the leading limb on the center of mass. Given the relationship between mechanical work and metabolic cost [3], it is possible that the elevated walking cost seen in children with cerebral palsy (CP) may be related to an altered amount of limb work performed during the single and double support phases. The purpose of this investigation was to evaluate the external individual limb work performed by children with CP during gait.

Statement of Clinical Significance

The results presented here provide further insight on the external limb work performed on the center of mass during the single and double support phases of children with CP. Insights from our results may help to guide the development of new rehabilitative strategies and biomechanical devices that are aimed at improving the walking mechanics of children with CP.

Methods

Ten children that were diagnosed as having CP with spastic diplegia (Age = 9.1 $\pm$ 2; Mass = 31.6 $\pm$ 13 kg), and ten healthy children with no walking disabilities (Age = 9.4 $\pm$ 2; Mass = 32.6 $\pm$ 10 kg) participated. There were no significant differences in the age (p=0.81) and mass (p=0.85) of the two experimental groups. None of the children with CP required walking assistive devices. Each participant walked at a self-selected pace along a 16-meter walkway. We collected individual limb ground reaction forces from four

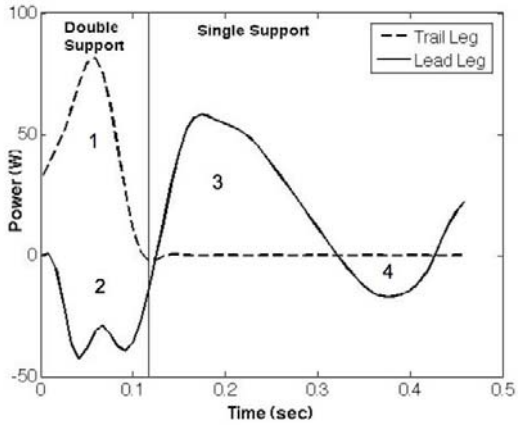

Figure 1. Exemplary limb power curves for a step from a healthy child walking at 1.3 m/s. Work was quantified as the area under the power curve. 1= the trail limb work during double support, 2= lead limb work during double support, 3= positive work during single support, and 4 = negative work during single support.

AMTI force platforms (120 Hz) that were mounted in series, and evaluated three steps from each participant. The individual limb ground reaction force components were summed to determine the acceleration of the center of mass. Each component of the acceleration was integrated to determine the instantaneous velocity of the center of mass [1]. We calculated the external mechanical power performed by the lead and trail limbs from the dot products of the respective individual limb forces and the velocity of the center of mass [3]. The amount of external work performed by each of the respective limbs was quantified by integrating the power curve over the time domain of interest. The work values were normalized by the participant's body mass. We evaluated the amount of positive and negative external work performed by the lead and trail limbs during the double and single support periods (Figure 1). Independent t-tests with a Bonferroni corrected alpha level (α=0.01) were used to determine if there were differences in the amount of individual limb work performed by the children with CP and controls during the respective periods of the step.

Results

The children with CP walked at a significantly slower speed than the controls (CP = 1.1 m/s; Controls = 1.3 m/s; p=0.005). However, there were significant differences in the amount of external limb work performed by the controls and children with CP (Figure 2). The children with CP performed significantly more negative work by the lead limb during double support (p=0.0004), and significantly less positive work by the trail limb (p< 0.00001). During single support, the children with CP performed significantly more positive work (p< 0.00001). No significant differences were found for the amount of negative work performed by the limb in single support (p = 0.84).

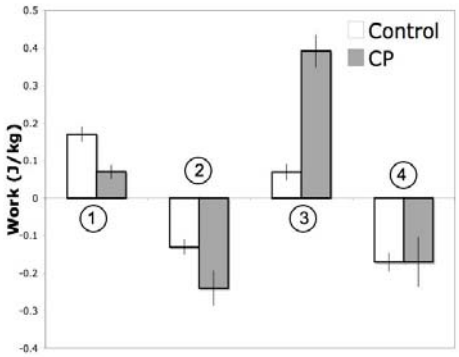

Figure 2. Mean (±SE) work performed by the lead limb during double support (1), trail limb during double support (2), positive work during single support (3), and negative work during single support (4).

Discussion

Our results indicate that children with spastic diplegic CP may lack the ability to appropriately perform external mechanical work to lift and redirect the center of mass. We speculate that the large amount of negative work that was performed during the double support phase may have influenced the amount of positive work that was performed during single support. In other words, the additional positive external work performed on the center of mass may have increased the body's potential energy during single support, and this energy was used to restore the energy lost during double support. It is possible that the amount of limb work performed during double support may have been influenced by the ankle-foot orthoses worn by many of the children with CP. This notion is supported by a recent investigation that has shown that an ankle-foot orthosis can influence the vertical excursion of the center of mass during gait [4]. We are currently investigating the influence of bracing on the amount of external mechanical work performed on the center of mass.

References

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Interrelationships among work, energy transfer, and the cost of walking in children

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Introduction

One hallmark of gait pathology is an elevated metabolic cost (MC) of walking. In an attempt to better understand the determinants of this poor locomotor economy, several investigators have studied the relationships between mechanical work (MW), mechanical energy transfer (ET), and MC [e.g., 2]. The results thus far have been equivocal, which may be partly due to limitations in the approaches used to compute MW and ET. Few investigators have computed MW from joint moments, and none have computed ET from joint moments using a model of the whole body combined with an induced power analysis [1, 4]. Therefore, the purpose of this study was to determine the interrelationships among MW, ET, and MC during walking in children, with both MW and ET computed from the net joint moments. We hypothesized that MC would be positively correlated with MW and negatively correlated with ET. We also evaluated the relationship between MW and ET.

Statement of clinical significance

Interpretation of gait analysis and metabolic data is a key component of the clinical decision-making process in the treatment of gait pathology. This study seeks to better establish the relationships between mechanical and metabolic energy utilization in able-bodied children.

Methods

Kinematic, kinetic, and oxygen consumption measurements were made on 18 able-bodied children over the age of eight (M±SD: age 11.9±2.2 yr, mass 45.7±12.8 kg, height 1.52±0.13 m). The gait data were processed using EvaRT 5.0 and OrthoTrak 6.3.4 to compute hip, knee, and ankle joint powers (joint moment × angular velocity). The bilateral joint powers were rectified, integrated over the gait cycle, and then summed across joints to yield an expression for MW. The gait data also served as inputs to a simulation model [5] that was used to perform an induced power analysis [1], which yielded the contribution of each joint moment to the segment powers for the ipsilateral limb, contralateral limb, and head-arms-trunk segment. From these data, an expression for the ET between segments was obtained. The oxygen consumption data were converted to the equivalent metabolic energy consumption, and then scaled by body mass and walking speed to represent the MC of walking. In order to make all of the variables dimensionally consistent, MW and ET were also scaled by body mass and walking speed. The interrelationships among MW, ET, and MC were evaluated using bivariate correlation coefficients and multiple linear regression.

Results

MC was positively correlated with MW (Fig. 1A) and negatively correlated with ET (Fig. 1B). ET was also negatively correlated with MW (Fig. 1C). The regression model was significant ($p < 0.001$, $r^2 = 0.64$), and the resulting equation was MC = 3.99 + 0.36 MW – 0.19 ET. The MW term was significant ($p = 0.005$), but the ET term was not ($p = 0.308$).

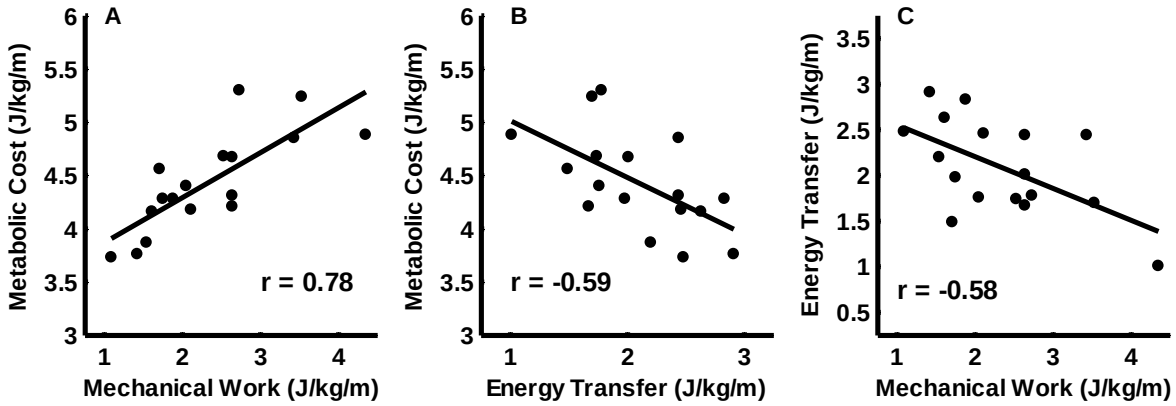

Fig. 1. Interrelationships among MW, ET, and MC during walking in children.

Discussion

The results supported our hypotheses that MC would be positively correlated with MW and negatively correlated with ET. However, ET was not a significant predictor of MC in the regression model. This is partly explained by greater variability in the MC-ET relation (compare Fig. 1B with 1A), but is also due to the collinearity of MW and ET (Fig. 1C). Together, these data reveal the interesting result that subjects who have limited ET between body segments while walking perform more MW, and have a higher MC. In our sample, it tended to be the youngest children (8-10 years) who had the least ET, performed the most MW, and had the highest MC. The present analyses were based on gross MC, which may not be the best representation of the true locomotor costs. To address this, the analyses were repeated using net non-dimensional cost [3] and non-dimensional MW and ET. These results were more variable, due primarily to variability in resting oxygen consumption, but showed the exact same trends as the gross MC data. Thus, it appears that as children get older they have more effective ET, which results in less of a need to perform MW, and ultimately leads to improved locomotor economy. This conclusion should be viewed in light of the magnitudes of the correlation coefficient values (Fig. 1), which imply that there must be other factors besides MW and ET which help determine the MC of walking. Utilization of stored elastic energy and co-contraction of antagonist muscle groups are two factors that could influence MC [2], but are not accounted for in the present model.

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Acknowledgements

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KINEMATICS, METABOLIC COST AND EXTERNAL WORK DONE DURING WALKING WITH A FORWARD ASSISTIVE FORCE

Introduction

It is well understood that many factors can change the metabolic cost of walking although the biomechanical mechanism is sometimes unclear. Gottschall and Kram [1] demonstrated that by applying a horizontal force at the center of mass of a healthy adult, the metabolic cost of walking can be reduced. We explored how an assist force changes gait kinematics and the external work that accompany the reduced metabolic cost in order to better understand this process.

Clinical Significance

The use of an assist force to reduce the metabolic cost of walking could result in development of therapeutic interventions for the elderly, children with cerebral palsy, or other individuals with disabilities who have an increased metabolic cost of walking or decreased endurance.

Methods

Subjects performed five walking trials on a level treadmill at their self-selected comfortable walking speed (CWS) while kinematic and metabolic data were collected. Subjects wore a full body marker set of 38 markers and a facemask to measure gas exchange. Three-dimensional kinematic data were collected using an 8-camera Vicon Motion Analysis System at 120 Hz while metabolic data was collected via the Oxycon Mobile wireless portable ergospirometry system.

Subjects were a convenience sample of 10 healthy adult males without known musculoskeletal, neurological, cardiac, or pulmonary pathology [age = 32.8±13.2, height = 1.8±.07 m, mass = 77.5±12.0 kg]. Consent was approved and obtained for all subjects. In all trials, steady state metabolic rate was determined by monitoring real-time VO2/kg consumption until two minutes of constant rates were observed. Subjects performed an unassisted “warm up” trial, three trials in random order with horizontal assistive forces applied equal to 4%, 8%, and 12% of their body weight, and a final “cool down” unassisted trial. Each trial was preceded by a rest period long enough for resting steady state metabolic rate to be reached. Constant assist forces were applied through a waist belt strapped roughly around the subject’s center of mass. After the experiment, subjects filled out a brief questionnaire asking about the experience of walking with an assist force.

Kinematic data were exported from Vicon and analyzed in MATLAB to find stride length, time spent in double support, recovery factor, relative phase, external work (W_{ext}), and potential to kinetic energy ratio (PE/KE) as described by Bennett et al [2]. Center of mass (COM) data was filtered with a bidirectional zero-lag low-pass second order Butterworth filter with a cutoff frequency of 8 Hz. Differences between measures were determined using repeated measures ANOVA with Tukey’s post hoc test. No significant differences were found between “warm up” and “cool down” for any measures, so data from the two were averaged to give the “normal” condition.

Results

During assisted trials, the metabolic cost of walking at CWS was shown to reduce by on average 23.7±6.3% of normal walking for 4% assistance force, 34.7±6.9% for 8% assistance, and 19.1±12.8% for 12% assistance ($p > .001$ for all). Stride length was found to decrease as assist force increased ($p<.026$). There was a trend of decreasing time in double support with increasing assist force. Hip excursion and ankle excursion were both found to decrease while knee excursion increased with increasing assist force ($p < .01$).

Table 1: Summary of metabolic and kinematic data.

	Normal	4%	8%	12%
% Reduction in Metabolic Cost	0.00%	23.70%*	34.66%*	19.05%*
Stride Length (m)	1.43	1.38*	1.35*	1.30*
% of Stride in Double Support	18.49%	17.48%	16.69%*	15.97%*
Recovery Factor	62.1	67.9*	67.6	63.0
Relative Phase (deg)	158.3	181.3*	199.4*	213.8*
W_{ext} ($J * kg^{-1} * m^{-1}$)	0.69	0.64	0.73	0.87*
PE/KE	1.16	1.13	1.13	1.12
Hip Excursion (deg)	47.6	43.5*	40.4*	35.5*
Knee Excursion (deg)	66.8	69.6*	71.5*	72.5*
Ankle Excursion (deg)	32.1	29.5*	26.7*	25.0*

* $p < .05$ compared to normal

Relative phasing between potential and kinetic energy of the COM during walking showed that phase angle increased as assist force increased ($p<.005$). Recovery factor improved slightly at 4% ($p=.044$) and 8% ($p=.058$) assist force but not at 12% ($p=.972$). There were no significant differences in the PE/KE ratio between trials, although excursion of kinetic energy and potential energy both increased.

Four of ten subjects reported preferring 4% assist, five preferred normal walking, while only one preferred 8%. Nine of ten subjects reported 12% as being the most uncomfortable trial.

Discussion

Our measured reductions in metabolic cost are similar to but less than those found by Gottschall and Kram [1]. This could be because their subjects performed a standing rest while ours rested seated and/or because our subjects walked at CWS rather than a uniform speed. The assist forces changed the kinematics of gait and the work on the CoM very little. We found 8% assist to be the most metabolically efficient condition although 4% assist had the highest recovery factor and best phasing between PE and KE. 8% assist also showed an increase in W_{ext} despite the decrease in metabolic cost. Thus is appears the assist forces were contributing to the work of changing the kinetic energy by providing propulsion assistance as well as changing the potential energy by raising the COM as it pivots over the stance leg, rather than changing the gait pattern.

References

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Acknowledgements

We would like to thank Shawn Russell for his assistance and advice.

EFFECT OF WALKING SPEED ON MECHANICAL RECOVERY AND INTERNAL WORK AFTER STROKE

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Introduction

Post-stroke hemiparetic gait is characterized by slower self-selected walking speed (SSWS) and asymmetric movement patterns (1,2). In the sagittal plane, gait can be modeled as an inverted pendulum, with energy being exchanged in order to alternately raise the center of mass and propel it forward. The Recovery Index is a measure that reflects the completeness of that energy exchange. In healthy ambulators, optimal percent recovery (~65%) occurs near the SSWS. Furthermore, the internal work done by the limbs (W_{int}) with respect to the center of mass increases with increased walking speed (3). It is unknown how decreased walking speed in stroke survivors affects percent recovery and internal work. This study investigates the effect of walking speed on the percent recovery and internal work in healthy and stroke subjects in order to evaluate changes in movement patterns.

Clinical Significance

Analyses of hemiparetic gait may help reveal sources of atypical mechanical energy expenditure during portions of the gait cycle and across speeds (1,4). Stroke subjects may use compensations or strategies in order to swing the paretic limb during ambulation which could lead to increased mechanical work and decreased percent recovery. By calculating the effect of increased walking speed on percent recovery and internal work, the effectiveness of stroke rehabilitation modalities that aim to increase SSWS could be evaluated.

Methods

Seven healthy young adults (age: 19-38 years, weight: 60-88 kg) and eight older stroke survivors (age: 49-78 years, weight: 57-104 kg) gave informed consent approved by Institutional Review Board. Stroke subjects were excluded if they had multiple strokes affecting both sides of the body. An over ground walking trial was conducted in order to determine each subject's SSWS. Twenty-two reflective markers were placed on bony anatomical landmarks of the foot, shank, thigh, pelvis, and trunk. Healthy subjects walked on a treadmill for twenty seconds at six different speeds ranging from 20% to 200% of their SSWS. Stroke subjects walked on a treadmill for twenty seconds at five different speeds ranging from 20% slower than their SSWS to the fastest speed at which they could safely walk while holding on to a handrail. Kinematic data were collected on all subjects using 6 infrared cameras and processed in Visual 3D in order calculate the instantaneous total mechanical energy of the whole-body center of mass: $E_{cm}=\frac{1}{2}mgh+\frac{1}{2}mv_{ap}^2+\frac{1}{2}mv_v^2$. External work ($W_{ext}$) was calculated by summing the positive increments of the instantaneous total mechanical energy curve (E_{cm}). Work in the vertical direction (W_v) and work in the anterior-posterior direction (W_{ap}) were calculated by summing the positive increments of the vertical and a-p components of the E_{cm} curve.

Percent recovery was calculated: $R=[(W_v+W_{ap}-W_{ext})/(W_v+W_{ap})]*100\%$. The instantaneous kinetic energy of each of the lower limb segments with respect to the whole-body center of mass was calculated: $KE=\frac{1}{2}mv_v^2+\frac{1}{2}mv_{ap}^2+\frac{1}{2}I\omega^2$. Energy transfers were assumed to occur between segments of the same limb but not between limbs. W_{int} was calculated by summing the positive increments of the kinetic energy curve for each limb. W_{int} was normalized to body mass (kg) and distance walked (m) which was calculated by multiplying walking speed (m/s) and trial duration (s).

Results

Average SSWS of stroke subjects (0.72 ± 0.14 m/s) was slower than SSWS of healthy subjects (1.15 ± 0.13 m/s). For healthy subjects, percent recovery reached a maximum (61% - 77%) when walking speed was between 75% and 125% of SSWS (Figure 1A). For five stroke subjects, percent recovery peaked at speeds above SSWS. For healthy and stroke subjects, internal work increased steadily with increased walking speeds (Figure 1B). Most stroke subjects exceeded the internal work done by healthy subjects.

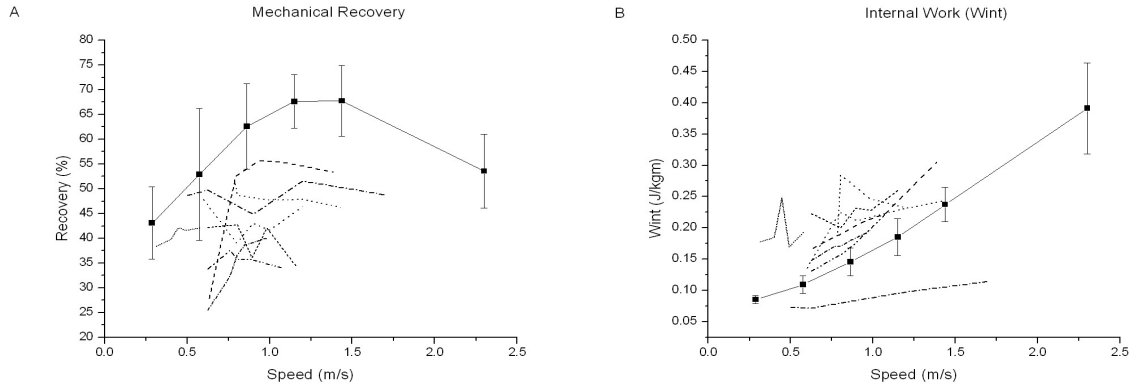

Figure 1: A. Average healthy recovery represented by solid line with ± 1 SD error bars. All other lines represent individual recovery values for each stroke subject. **B.** Average healthy internal work represented by solid line with ± 1 SD error bars. All other lines represent individual internal work values for each stroke subject

Discussion

Maximum percent recovery is expected to occur near the SSWS in healthy subjects (3). Most stroke subjects improve their percent recovery with increased walking speed above SSWS while also experiencing the same increases in internal work as healthy subjects. Our results suggest that gait retraining targeted at increasing SSWS may improve recovery of mechanical energy while not causing additional increases in internal work due to compensation strategies.

References

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Acknowledgments

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Muscle contributions to propulsion and support during running

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INTRODUCTION: Understanding how muscles propel our body forward and support our body weight during running is challenging because important variables, such as muscle forces, are generally not measurable. Muscle-actuated simulations of walking have been analyzed to gain insight into muscle actions[1,2]; however, simulating running presents new challenges due to high accelerations and forces. The purpose of this study was to create a three-dimensional, muscle-actuated simulation of running and quantify how muscles contribute to propulsion and support (i.e., the forward and vertical accelerations of the body mass center) during running.

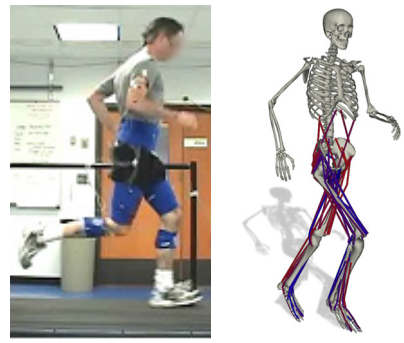

Figure 1. A three-dimensional, muscle-actuated simulation of the running gait cycle generated from motion capture.

CLINICAL SIGNIFICANCE: Running has become a popular activity, with at least 10.5 million Americans running 100 days or more in 2002[3]. With a large population of runners there is also large potential for injury, since every year 25-50% of runners sustain an injury[4]. Understanding muscle forces and the motions they produce during running could help coaches and physical therapists improve training and rehabilitative techniques[5] to prevent common injuries, such as acute hamstring injury due to large muscle strain during foot strike[6].

METHODS AND PROCEDURES: Kinematic and ground reaction force data were measured on an adult male subject running at 3.9 m/s on an instrumented treadmill (Fig. 1). The subject was represented by a 31 degree-of-freedom musculoskeletal model with the lower extremities and back driven by 92 musculotendon actuators[7,8] and upper extremities[9] driven by torque actuators (Fig. 1). OpenSim[10] was used to generate a simulation of the muscle activations and forces needed to drive the model to track the experimentally measured motion.

A perturbation analysis[11] was performed to compute contributions of individual muscles to propulsion (the forward acceleration of the body mass center) and support (the upward acceleration of the body mass center). At each time step in the simulation, each musculotendon actuator was individually perturbed by increasing its force by 1 N (ΔF_m). The perturbed model was then integrated forward 0.02 s (Δt), and the change in mass center acceleration due to the perturbation was calculated by comparing the perturbed ($\boldsymbol{x}_{perturbed}$) and unperturbed ($\boldsymbol{x}_{unperturbed}$) mass center locations (Eq. 1). This change in acceleration ($\partial \ddot{\boldsymbol{x}} / \partial F_m$) represents a muscle's potential to accelerate the mass center. The potential was then multiplied by the corresponding unperturbed muscle force ($F_{unperturbed}$) from the simulation to calculate the acceleration contributed by that muscle (Eq. 2).

$$\frac{\partial \ddot{\boldsymbol{x}}}{\partial F_m} = 2 \cdot \frac{\boldsymbol{x}_{perturbed} - \boldsymbol{x}_{unperturbed}}{\Delta t^2 \cdot \Delta F_m} \quad \text{(Eq. 1)}$$

$$\ddot{\boldsymbol{x}}_m(t_i) = \frac{\partial \ddot{\boldsymbol{x}}}{\partial F_m} \cdot F_{unperturbed} \quad \text{(Eq. 2)}$$

Our analysis focused on the propulsive phase of stance, defined as the time period when a foot is on the ground and the body mass center is accelerating forward.

RESULTS: Gastrocnemius, soleus, and hamstrings contributed most to propulsion, with average forward accelerations of 1.4, 1.0, and 0.9 m/s^2, respectively, during the propulsive phase of stance (Fig. 2). In contrast, the quadriceps muscle group resisted forward propulsion, with an average contribution of -2.7 m/s^2. Soleus, gastrocnemius, and quadriceps contributed the most to vertical support, with average contributions of 3.3, 2.7, and 2.0 m/s^2, respectively. Hamstrings acted to pull the body mass center downward, with a contribution of -0.7 m/s^2.

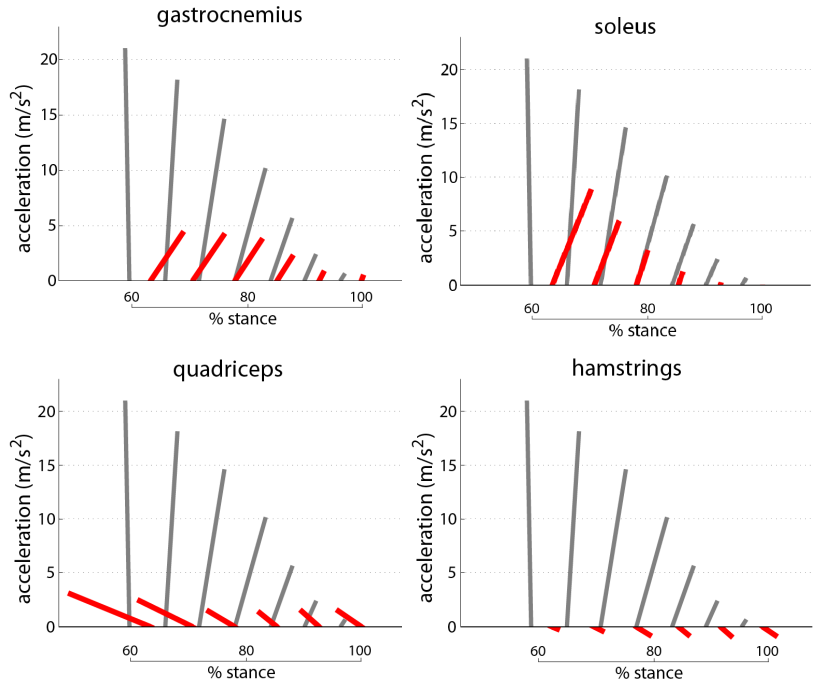

Figure 2. Accelerations of the body mass center from gastrocnemius, soleus, quadriceps, and hamstrings (red vectors). Vectors represent the resultant of forward and upward accelerations of the mass center. Gray vectors represent the total acceleration of the body mass center (i.e., the ground reaction force divided by total body mass).

DISCUSSION: The ankle plantar-flexors are a key muscle group for both forward propulsion and upward support of the body mass center during the propulsive phase of running. The gastrocnemius contributes more to propulsion and soleus contributes more to support, which is consistent with results of previous studies[4,12].

Having a three dimensional model with arms also allowed us to investigate the contribution of arm motion to forward propulsion, vertical support, and medial-lateral movement. Hinrichs[13] concluded that arms provide lift (i.e., support) and do not contribute to forward propulsion. Our results suggest, for this subject, arm motion did not contribute substantially to acceleration of the mass center in any direction. Further analysis is needed to investigate if the arms act to counter angular accelerations of the legs, as suggested by Novacheck[4].

We have used the first three-dimensional, muscle-driven simulation of a running gait cycle to answer the fundamental question of which muscles propel the body forward and support body weight during running. This analysis is for one subject at one speed on a treadmill, and additional simulations are needed to investigate how muscle contributions change with running speed, in different conditions (i.e., over-ground vs. treadmill), and vary between subjects. With broader understanding of muscle actions during running, individual simulations could then give more insight into specific training techniques for injury prevention.

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GAIT CLASSIFICATION DEPENDS ON WALKING SPEED

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Introduction

Gait classifications are generally considered to be important for understanding gait deviations in children with cerebral palsy, as well as for diagnosis, clinical decision making and communication. Several classification systems have been proposed and used throughout the literature [1]. Although exact definitions vary, typical classifications include crouch gait (excessive knee flexion in stance), hyperextended gait (excessive knee extension in stance), and equinus gait (excessive ankle plantar flexion or toe walking), or a combination of those. The aim of this study was to determine whether these gait classifications are a consistent patient characteristic or if (and if so, how) they change with walking speed.

Clinical Significance

Walking speed can vary considerably between patients and controls, between pre- and post treatment, and even between trials within one session. This study leads to a better understanding of the effects of walking speed, which is necessary for comparing gait analyses at varying speeds.

Methods

17 children with spastic CP (aged 6-12) and 11 matched typically developing (TD) children participated in this study. Each subject walked at 70(±5)%, 100% and 130(±5)% of comfortable walking speed (CWS). Speed was recorded online, and controlled by giving verbal feedback after each trial. No restrictions were imposed on step length or frequency. Six successful trials were performed for each walking speed, divided over two sessions. 3D kinematic data were collected (OptoTrak, NDI), and analyzed according to ISB standards [2]. Norm values were calculated as the mean ± 2SD of the TD subjects at CWS. Gait patterns were classified based on knee and foot angles in stance, as follows:

- **Crouch**: peak knee extension during mid-terminal stance (30-60 %Gait Cycle) < Norm;
- **Hyperextension**: peak knee extension during mid-terminal stance (30-60 %GC) > Norm;
- **Equinus**: toe walking or early heel rise, mean plantar flexion of the foot (relative to global) during mid stance (20-40 %GC) > Norm;
- **Neutral**: no crouch, hyperextension or equinus present

Results

All children were able to walk at the imposed range of speeds without problems. At CWS, all TD legs were classified as neutral; 9 CP legs as crouch; 11 as knee hyperextension and 27 as equinus. The gait classification changed with walking speed for 19 of 33 CP legs and for 9 of 11 TD legs (Table 1). The main changes were that at faster speeds many CP legs developed hyperextension (8/33 legs, fig. 1A) and equinus gait (9/33 legs, fig. 1B) that were absent at slower speeds. 4 legs went from crouch to no crouch. TD legs tended to be in crouch at slow speed (3/11 legs) and in knee hyperextension at fast speed (4/11 legs).

Table 1. Gait classifications at slow, comfortable and fast walking speed for each limb

SUBJ	CP RIGHT LEG			CP LEFT LEG			TD RIGHT LEG		
	SLOW	CWS	FAST	SLOW	CWS	FAST	SLOW	CWS	FAST
1	--	--	--	CE	CE	CE	N	N	N
2	C	C	CE	H	H	N	N	N	H
3	E	HE	HE	E	E	E	N	N	HE
4	E	HE	HE	HE	HE	HE	N	N	E
5	E	HE	HE	N	HE	HE	N	N	N
6	HE	E	E	E	E	E	C	N	C
7	CE	E	E	E	E	HE	C	N	N
8	HE	HE	HE	CE	CE	CE	N	N	H
9	CE	E	--	E	E	HE	C	N	N
10	E	E	E	E	HE	HE	N	N	H
11	N	N	N	H	H	HE	N	N	N
12	C	CE	E	C	E	E			
13	CE	CE	CE	CE	CE	CE			
14	C	C	CE	CE	CE	CE			
15	N	HE	H	HE	HE	HE			
16	N	E	E	N	N	E			
17	E	E	E	CE	CE	CE			

C = Crouch; H = Hyperextension; E = Equinus; N = Neutral; -- = missing for technical reasons
CP = cerebral palsy; TD = typically developing; CWS = comfortable walking speed

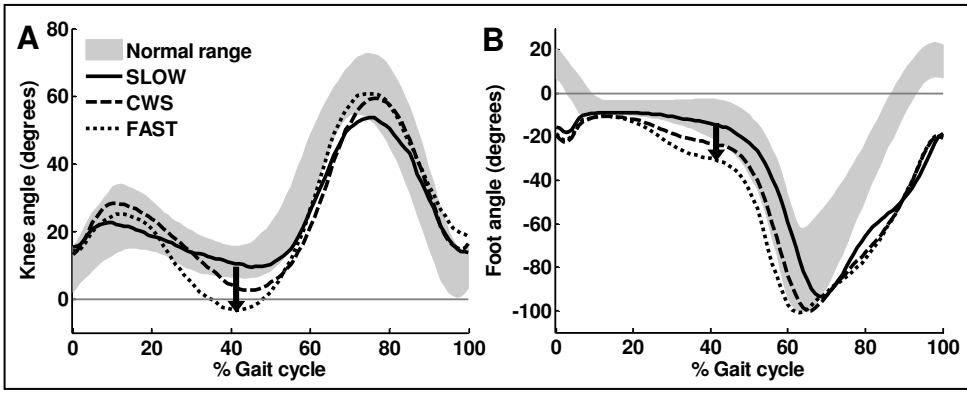

Figure 1. Example knee (A) and foot (B) angle data for slow, comfortable (CWS) and fast walking speed for two typical subjects going from neutral to A) hyper-extension and B) equinus.

Discussion

This study shows that gait classifications can vary with walking speed and are not a fixed patient characteristic. Knee (hyper)extension and equinus tend to increase with speed. Even typically developing subjects walk in 'pathological' gait patterns when walking slow (mostly 'crouch gait') or fast (mostly 'knee hyperextension'). These effects of walking speed are important to consider when comparing clinical gait data at different speeds, and may give insight into why comfortable walking speed is reduced. Furthermore, the results indicate that gait classification should be interpreted with caution. Gait classification systems typically divide a continuum of kinematic data into groups. Whether a patient falls inside or outside of a group can depend on small changes in kinematics. Our results call for a gait identification system that not just classifies, but also grades gait patterns, based on the main characteristics of pathological walking.

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Crouch Gait Patterns Derived from Cluster Analysis are Related to Clinical Parameters and Surgical Interventions

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INTRODUCTION

Classification schemes have been used to describe walking patterns of children with cerebral palsy (CP), and to help guide treatment decisions [1]. One such pattern is crouch gait, which is characterized primarily by increased knee flexion in stance. A method has been developed to automatically identify homogeneous groups within the overall pattern of crouch gait [2]. The method uses a reduced order approximation of gait based on a limited number of *gait features* – elements of an optimal basis for describing kinematics. A subject's gait can be well described by just a few gait feature scores. When the gait feature scores of subjects in crouch are organized using a k-means cluster analysis, homogeneous groups that relate to underlying pathology emerge [3]. The present study explores the relationship between group membership and clinical parameters such as strength, selective motor control, and spasticity.

CLINCAL SIGNIFICANCE

Delineating meaningful groups within the overall gait pattern known as crouch will allow clinicians to better focus their treatment plans.

METHODS

Using singular value decomposition, gait features of sagittal plane kinematics were derived from a large database of strides [2]. The first four gait feature scores were calculated for children with Cerebral Palsy who walked with >20º knee flexion at initial contact. K-means cluster analysis was performed on the gait feature scores and the resulting clusters were numbered by increasing pathology (1-mildest, 5-most severe) as measured by their proximity to normal gait [2].

Three clinical domains were then used to characterize the subjects' underlying pathology:

- **Strength** at the hip, knee, and ankle:
  Kendall strength score [1 (weak) through 5 (strong), 3 = anti-gravity]
- **Selective motor control** of the hip, knee and ankle:
  movement [0 = patterned, 1 = partially isolated, 2 = completely isolated]
- **Spasticity** of the hip flexors, hamstrings, rectus femoris, and plantarflexors:
  modified Ashworth score [1 (no spasticity) through 5 (rigid)]

Strength was highly correlated between the hip, knee and ankle, as were selective motor control and spasticity. Because of this, principal component analysis was used to derive a single overall measure for each of these clinical domains. The first principal component of strength, selective motor control, and spasticity was used as the overall measure (accounting for 64%, 58%, and 52% variance respectively).

RESULTS

The knee hyper-flexion criterion was met by 2956 strides. K-means cluster analysis resulted in five crisp clusters with robust centers. Based on the observed gait patterns the clusters were named, in order of increasing severity, 1) mild crouch + mild equinus, 2) crouch, 3) crouch + anterior pelvic tilt, 4) crouch + equinus, and 5) severe crouch.

A one-way ANOVA showed significant differences in strength, spasticity, and selective motor control amongst the five clusters [*e.g.* Figure 1]. Of the 2956 strides in crouch, 872 were from subjects who underwent a subsequent gait analysis, 746 of whom had intervening surgery. The distribution of these surgeries was also observed to differ markedly between the crouch groups [*e.g.* Figure 2].

DISCUSSION

The results of this study show that the various crouch groups are characterized by different underlying clinical parameters. For example, group 3 (crouch + anterior pelvic tilt) exhibits fairly mild spasticity while group 4 (crouch + equinus) has severe spasticity. Related to this is the finding that, among other differences, group 4 subjects are much more likely to undergo a selective dorsal rhizotomy. The crouch groups defined using the gait features and k-means cluster analysis arise from purely mathematical considerations. However, the strong relationship of these clusters to underlying pathology and surgical intervention suggest that they may represent a valuable scheme for categorizing subjects who walk in crouch.

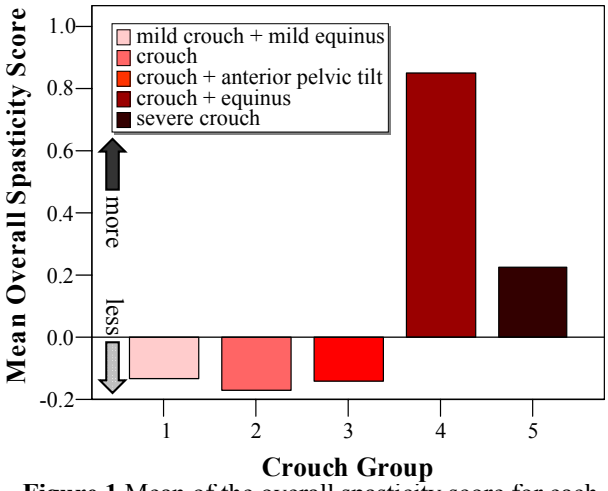

Figure 1 Mean of the overall spasticity score for each crouch group. A higher number indicates more spasticity. Note the large difference between groups 3 and 4.

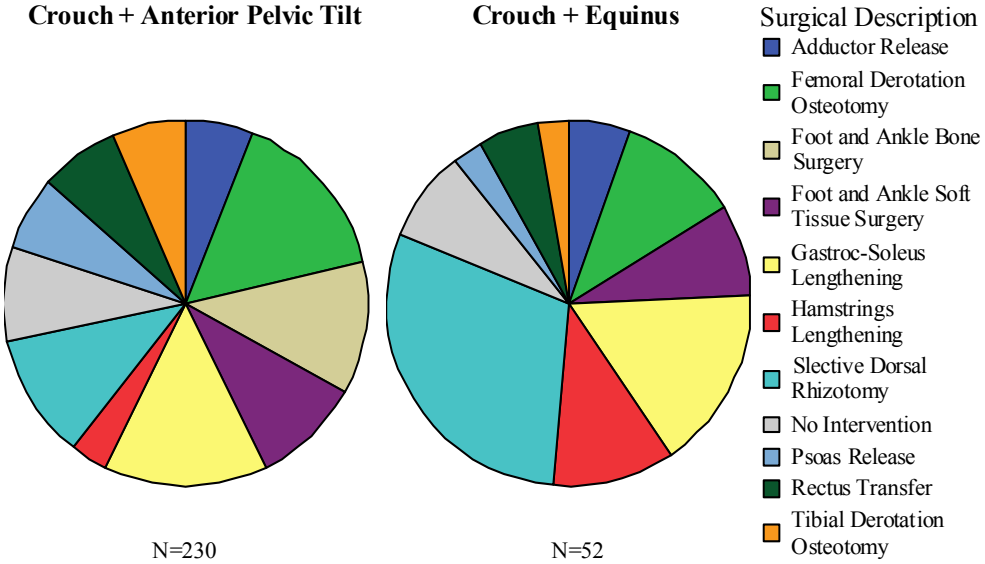

Figure 2 Example of the distribution of surgeries among crouch groups. The crouch + equinus group has a higher incidence of selective dorsal rhizotomy and also has the highest spasticity.

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INFLUENCE OF GAIT ANALYSIS ON DECISION MAKING FOR HAMSTRING LENGTHENING SURGERY

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INTRODUCTION

Several studies have shown that surgical decision making is altered when gait analysis data are included in the decision making process[1-4]. However, a weakness of these previous studies is that they have been performed on observational cohorts, which makes it difficult to isolate the effects of gait analysis and to control for differences among patients and surgeons. This study used data from a randomized, controlled trial (RCT) to examine the influence of gait analysis data on decision making for hamstring lengthening surgery (HSL).

Our main hypothesis was that decision making for HSL surgery would be influenced by recommendations received in a gait analysis report. Specifically, we hypothesized that: (1) Surgeons are more likely to proceed with planned HSL surgery if they receive a gait report supporting this plan; (2) Surgeons are more likely not to proceed with planned HSL surgery if they receive a gait report recommending against this plan; (3) If HSL is not planned initially, it will be performed more often when the surgeon receives a gait report recommending HSL; (4) If HSL is not planned and the gait report agrees that it should not be done, receiving a gait report will have no effect because HSL will rarely be done.

CLINICAL SIGNIFICANCE

These results from a randomized, controlled trial provide a stronger level of evidence demonstrating the impact of clinical gait analysis on surgical decision making.

METHODS

This study included 130 ambulatory children with cerebral palsy (CP), age 3-18 yr, who were candidates for lower extremity orthopaedic surgery to improve gait. All subjects underwent pre-operative gait analysis including physical examination, computerized gait analysis, and electromyography, and standard clinical gait reports were produced. The subjects were randomized to two groups: 1) Treatment group, where the referring surgeon received the patient's gait analysis report and 2) Control group, where the referring surgeon did not receive the gait report.

Data on HSL and other surgeries were collected at three time points: 1) referral by treating surgeon before gait analysis, 2) recommendations by gait laboratory surgeon after gait analysis, and 3) actual surgery performed. The unit of analysis was patient-side. For unilaterally involved subjects, only the affected side was included. The main outcome measure was the relative acceptance between the Treatment and Control groups, where acceptance is defined as the likelihood that the referring surgeon accepted the gait analysis recommendation for HSL. Statistical significance was determined using the 2-sided Fisher's exact test.

RESULTS

Of those sides that had HSL planned by the referring surgeon before gait analysis, the vast majority also had HSL recommended by the gait laboratory (*Both referral & gait lab*) (Table 1). HSL was ultimately performed in 90% (19/21) of cases in the Treatment group (after review of the gait report), compared with 63% (19/30) of cases in the Control group (without access to the gait report), relative acceptance: 1.43 (95% CI: 1.05, 1.94; p=0.048). These results suggest that the gait reports reinforced the referring surgeons' original decision making, increasing their confidence to proceed with the surgery.

The gait reports also influenced decision making in the few cases where the gait laboratory disagreed with the referring surgeon's plan to do HSL (*Referral only*). The treatment plan was changed for the only side in patients receiving the gait report (1/1), but not for those without access to the gait report (0/4) (relative acceptance undefined; p=0.200).

The gait laboratory recommended HSL in a large number of cases where HSL was not planned before gait analysis (*Gait lab only*). HSL was performed in 14% (7/51) of sides in subjects who received the gait report, compared with 0% (0/34) of those not receiving the report (relative acceptance: undefined; p=0.038).

For those sides where the surgeons did not plan and the gait analysis did not recommend HSL (*None*), HSL was not performed in either group.

Table 1: HSL surgery at each time point for Treatment and Control groups

Planned Before GA ***22/108 (20%)*** 34/110 (31%)		***Treatment Group (N=108)*** Control Group (N=110)		Not Planned Before GA ***86/108 (80%)*** 76/110 (69%)			
Both Referral & Gait Lab		*Referral Only*		*Gait Lab Only*		*None*	
Recommended by GA ***21/22 (95%)*** 30/34 (88%)		Not Recommended by GA ***1/22 (5%)*** 4/34 (12%)		Recommended by GA ***51/86 (59%)*** 34/76 (45%)		Not Recommended by GA ***35/86 (41%)*** 42/76 (55%)	
Done ***19/21 (90%)*** 19/30 (63%)	Not Done ***2/21 (10%)*** 11/30 (37%)	Done ***0/1 (0%)*** 4/4 (100%)	Not Done ***1/1 (100%)*** 0/4 (0%)	Done ***7/51 (14%)*** 0/34 (0%)	Not Done ***44/51 (86%)*** 32/32 (100%)	Done ***0/35 (0%)*** 0/42 (0%)	Not Done ***35/35 (100%)*** 42/42 (100%)

DISCUSSION

The preliminary results from this ongoing RCT support our hypotheses although statistical significance was not always achieved. As more data are collected, statistical power should increase. It appears that gait analysis not only alters decision making as previously reported[1-4], it also reinforces decision making when gait analysis agrees with the surgeon's original plan. The randomized design of this study provides a baseline against which the gait analysis results can be compared. In a companion study using the same methodology, we found similar results for hip adductor lengthening surgery.

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Predicting changes in knee flexion after hamstrings lengthening surgery

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Introduction: Although many patients with crouch gait exhibit an improvement in knee flexion kinematics after hamstrings lengthening surgery, these improvements are neither consistent nor predictable[1,2]. Clinicians can draw on data from a patient's physical exam and gait analysis to help select good candidates for surgery, but interpreting this information is challenging, and surgeons often recommend different treatments from the same set of information. In this study, we developed a statistical model to predict the likelihood that a patient's knee flexion would improve after a hamstring lengthening. Our model makes predictions based on available pre-operative data and was validated with a retrospective analysis of subjects treated for crouch gait.

Clinical Significance: A statistical model for predicting the outcome of hamstrings surgery could aid clinical decision-making by providing an objective assessment of the likelihood a patient will benefit from treatment.

Methods:

Subjects: We analyzed 73 children who received a hamstrings lengthening to treat their crouch gait. All subjects had a pre- and post-operative gait analysis and physical exam. We included subjects with concomitant lower-extremity musculoskeletal surgeries, but excluded subjects who had ever received a selective dorsal rhizotomy or baclofen pump, or had received any lower extremity surgery in the 12 months preceding the pre-operative gait analysis.

Outcome Measure: We categorized a subject as "improved" after surgery if he or she had at least 25% improvement in knee flexion at initial contact or was within 2 standard deviations of normal after surgery (less than 16° knee flexion) and "unimproved" otherwise.

Model Development: We selected a set of 30 potential predictive variables from the subjects' gait kinematics, kinetics, and physical exam that were significantly different between the "improved" and "unimproved" groups. We then tested subsets of 3 to 7 variables to find the subset with the highest prediction accuracy. The statistical model used in this study was a Naïve Bayes Classifier[3], which estimated the probability that each subject would be "improved" after surgery based on the selected subset of pre-operative variables.

Model Testing: To test the predictive strength of our statistical model we used 10-fold cross validation[3]. We divided the subjects into 10 sets, and then successively fit the model, using 9 of the 10 to train the model, and the remaining, reserved set of subjects to test the model's accuracy. This process gives an estimate of the model's accuracy in making predictions for a new group of subjects.

Table 1: Pre-Operative Predictive Variables

	Improved	Unimproved	p-value
Terminal Swing Hip Power (W/kg)	0.25 (0.32)	0.09 (0.39)	0.044
Minimum Knee Flexion Angle (°)	37 (15)	27 (10)	0.003
Normalized Pelvic Tilt Score	91 (9.2)	96 (6.4)	0.004
Hip Extension Strength (0-5 Scale)	3.0 (1.0)	3.7 (0.9)	0.035
Hip Abduction in Physical Exam (°)	51 (15)	57 (11)	0.097
Psoas Lengthening (Frequency)	33%	55%	0.059

Results: Using pre-operative data, we were able to predict whether subjects' knee flexion at initial contact would improve after a hamstrings lengthening with 80% cross-validation accuracy. The predictive variables chosen for the model are shown in Table 1, along with the means and standard deviations for the "improved" and "unimproved" groups and the p-value for a two-sample t-test. Figure 1 shows the predicted likelihood of each subject falling into the "improved" category using this statistical model, versus actual % improvement in knee flexion. The "improved" subjects lie to the right of the vertical dashed line, demonstrating at least a 25% improvement in knee flexion. The subjects that the statistical model predicted would improve, with a probability of at least 0.5, lie above the horizontal dashed line. Based on these criteria, 29 subjects were correctly classified as "improved" (*solid black*) and 32 subjects were correctly classified as "unimproved" (*solid gray*), while 5 subjects were misclassified as "improved" (*open black*) and 7 subjects were misclassified as "unimproved" (*open gray*).

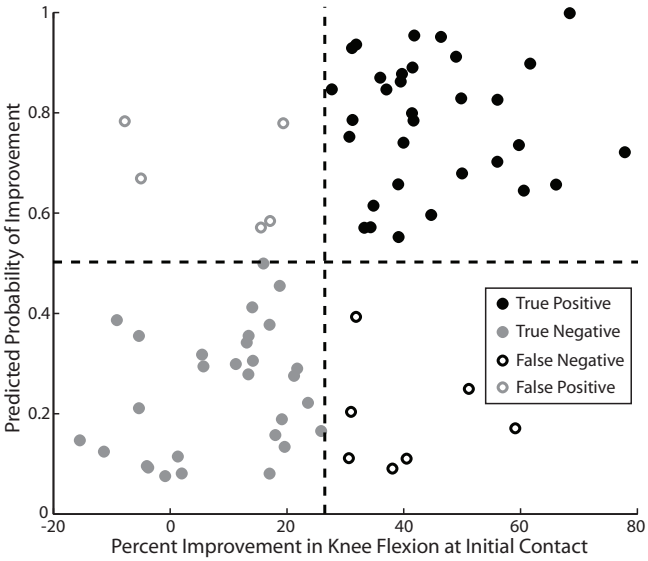

Figure 1: Predicted likelihood of each subject "improving" after surgery versus actual change in knee flexion

Discussion: The statistical model developed in this study correctly classified 80% of subjects as "improved" or "unimproved" after hamstrings surgery, in contrast to the actual improvement rate of 55%. Some of the predictive variables used in the model "make sense" in the context of our current clinical knowledge of crouch gait. For example, the "unimproved" subjects tended to have milder crouch, with minimum knee flexion angles closer to normal. In these subjects, the hamstrings may not restrict terminal swing knee extension and thus these subjects may not respond to a lengthening surgery. Interpreting some of the other predictive variables is more difficult. For example, kinematics at the pelvis was linked to improvement, while subjects who received a concomitant psoas lengthening were less likely to have improved knee flexion after hamstrings surgery. In some cases, these unexpected predictors may be surrogate variables for the factors that directly govern improvement after surgery. Indeed, the set of variables used in the model is not unique. Since many of the pre-operative predictive variables are correlated, several combinations of predictive variables achieved accuracies close to 80%. Further biomechanical investigation is needed to understand why variables like hip abduction range of motion and terminal swing hip power were important for predicting surgical outcome.

This study has demonstrated the potential benefit of using a statistical model to help identify good candidates for hamstrings surgery. The statistical model gives a set of weightings for each variable and an estimate of the probability that a subject's knee flexion will improve, both of which enhance the clinical accessibility of this approach. This model can also be modified to make predictions for other treatments, outcome metrics, or clinical centers.

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ACCURACY OF SIX DOF JOINT KINEMATICS DURING NORMAL AND PATHOLOGICAL GAIT: A SIMULATION STUDY

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INTRODUCTION

The conventional gait model (CGM) uses a minimal marker configuration to track three-dimensional lower extremity motion[1]. However, the CGM relies on shared markers between segments, which may adversely affect the reliability of non-sagittal joint angles[2]. A number of researchers have opted to use clusters of three or more markers on each segment to independently assess all 6 segmental degrees of freedom (DOF)[3]. A prior study by our group showed that gait measures derived using redundant marker sets differ significantly from those obtained using a CGM approach in normal subjects, particularly in the non-sagittal plane[4]. However, it is unknown whether the differences found actually reflect greater accuracy, or whether comparable differences would be observed in pathological gait. The goal of this study was to compare the accuracy of non-sagittal joint kinematics derived from inverse kinematic models based on shared and independent marker sets.

CLINICAL SIGNIFICANCE

Pathological gait often includes substantial non-sagittal motion. However, a lack of confidence in non-sagittal gait measures diminishes the usage of such data for treating movement disorders. The results of this study suggest that use of independent marker sets on each limb may improve the accuracy of transverse plane joint angles, and hence may allow for greater consideration of non-sagittal motion in treatment planning.

METHODS

We generated gait simulations of six healthy children (4 males, 2 females; age, 9 ± 1.5 yrs) and two cerebral palsy (CP) patients (1 male age 10 yrs with apparent equinus, 1 female age 14 yrs with Winters type 4 gait pattern) walking over ground at a preferred speed. To generate the simulations, we first scaled a generic three-dimensional, whole body model (with 34 degrees of freedom) to match the measured segment lengths of each subject. We then used a least squares forward dynamics algorithm to compute the pelvic motion and joint angular trajectories over a gait cycle that were optimally consistent with the measured marker kinematics and ground reactions, while also satisfying the overall equations of motion of the system[5]. The results were a set of joint torque-actuated forward dynamic simulations that closely resembled subject-specific gait patterns. Simulated angles were generally within 1 deg of those obtained using more conventional inverse kinematic approaches.

Ideal marker trajectories were then generated assuming that the markers were rigidly fixed to each segment in the simulations. Both random equipment noise (Gaussian noise with 1 mm s.d.) and soft tissue artefact (continuous noise model with random amplitude<10 mm, frequency f<4 Hz, and phase[6]) were then added to the noise free marker trajectories. We computed the joint angles from both the corrupted marker trajectories using six degree of

freedom inverse kinematic analyses with two marker sets: 1) A shared marker set in which a virtual hip joint center, and the lateral knee and ankle markers were shared by both the proximal and distal segments, 2) An independent marker set in which 4 tracking markers on each segment were used to compute 6 dof segmental position. We quantified accuracy as the root-mean-squared (RMS) difference between the reconstructed and actual joint angles over a full gait cycle.

RESULTS

Use of an independent marker set resulted in smaller RMS errors for hip, knee, and ankle internal/external rotation angles, and ankle ab/adduction when compared to a shared marker set (Table 1). However, errors in hip ab/adduction were slightly larger using an independent marker set. Model effects on accuracy were similar whether processing normal or pathological gait.

DISCUSSION

Our results suggest that the use of independent marker sets, in which 3 or more markers are placed on a rigid plate attached mid-segment, may enhance the accuracy of 3D joint angles. The most noteworthy improvements were internal-external rotation angles and likely result from using redundant non-collinear markers displaced from the long axis of the joint. Our analyses considered the effect of equipment noise and soft-tissue artefact. However, implementation also requires consideration of how soft tissue noise can vary systematically with marker placement. We conclude that the development and use of optimized, independent marker sets may enhance the confidence that practitioners have in using non-sagittal biomechanical measures to plan treatment.

Table 1: Root mean square (RMS) errors for non-sagittal joint angles using shared and independent marker sets on each segment.

Variable	Normal		Cerebral Palsy		
	Shared Marker Set	Independent Marker Set	Shared Marker Set	Independent Marker Set	Model Effect Significant?
Hip Ab/adduction	2.6 (0.7)	4.3 (1.0)	1.9 (0.8)	4.9 (0.3)	*
Hip Rotation	7.8 (3.7)	4.8 (1.5)	6.7 (2.1)	4.6 (1.6)	*
Knee Ab/adduction	3.3 (1.2)	4.4 (0.8)	4.9 (1.7)	5.2 (0.8)	
Knee Rotation	18.6 (7.0)	6.5 (1.6)	16.5 (4.1)	5.6 (0.6)	*
Ankle Ab/adduction	7.1 (2.6)	3.4 (0.9)	6.2 (1.3)	4.1 (0.6)	*
Ankle Rotation	16.0 (7.7)	4.8 (0.9)	13.9 (4.4)	4.0 (1.4)	*

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Accelerations produced by gastrocnemius and soleus during crouch gait

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Introduction

The ankle plantarflexors support body weight and propel the mass center forward during the latter half of the stance phase of unimpaired gait.[1,2] The gastrocnemius and soleus both provide support and propulsion by inducing plantarflexion accelerations at the ankle; however, these muscles have opposing functions at the knee. While the soleus accelerates the knee into extension via dynamic coupling, the bi-articular gastrocnemius has a knee flexion moment arm and weakly accelerates the knee into flexion during unimpaired gait[2,3]. In subjects with crouch gait, changes in joint kinematics and muscle activations may alter the accelerations generated by the plantarflexors. For example, the gastrocnemius exhibits abnormally high levels of activity in some subjects with crouch gait; this over-activation may contribute to the excess knee flexion or may be an effort to increase the COM support provided by the gastrocnemius. Understanding the effects of this over-activation requires that we establish the motions caused by the gastrocnemius and soleus during crouch gait. Thus, the goal of this study was to quantify the effects of the gastrocnemius and soleus on the acceleration of the ankle, knee, and hip and the body's COM in subjects with crouch gait.

Statement of Clinical Significance

To understand which muscles contribute to crouch gait and the effects of surgically lengthening the gastrocnemius and/or soleus a comprehensive understanding of their role in accelerating the joints and the COM during crouch gait is required.

Methods

We developed subject-specific dynamic simulations of 4 subjects [3 male, 1 female; age: 8.1(0.9) years; height: 125.6(4.6) m; weight: 32.1(12.6) kg] from Gillette Children's Specialty Healthcare who walked in a moderate crouch gait (knee flexion > 20° throughout stance) and had received no prior surgery. For each subject, gait data was imported into OpenSim and muscle forces were predicted for the single limb stance phase of four trials for each subject[4]. A musculoskeletal model with 19 degrees of freedom and 92 muscles was scaled to each subject. Muscle forces were predicted using the Computed Muscle Control (CMC) algorithm[5]. An induced acceleration analysis was then used to determine the contribution of each muscle to joint and COM accelerations. COM accelerations in the upward direction indicate a contribution to support while COM accelerations in the forward direction indicate a contribution to progression. We normalized each muscle's contribution to joint acceleration by the magnitude of the muscle with the largest induced acceleration at each joint. The muscles used for normalization included gluteus medius, vasti, and soleus, the muscles that made the largest contribution to accelerations at the hip, knee, and ankle respectively.

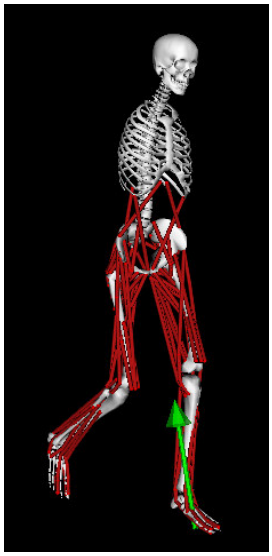

Figure 1: OpenSim model of crouch gait.

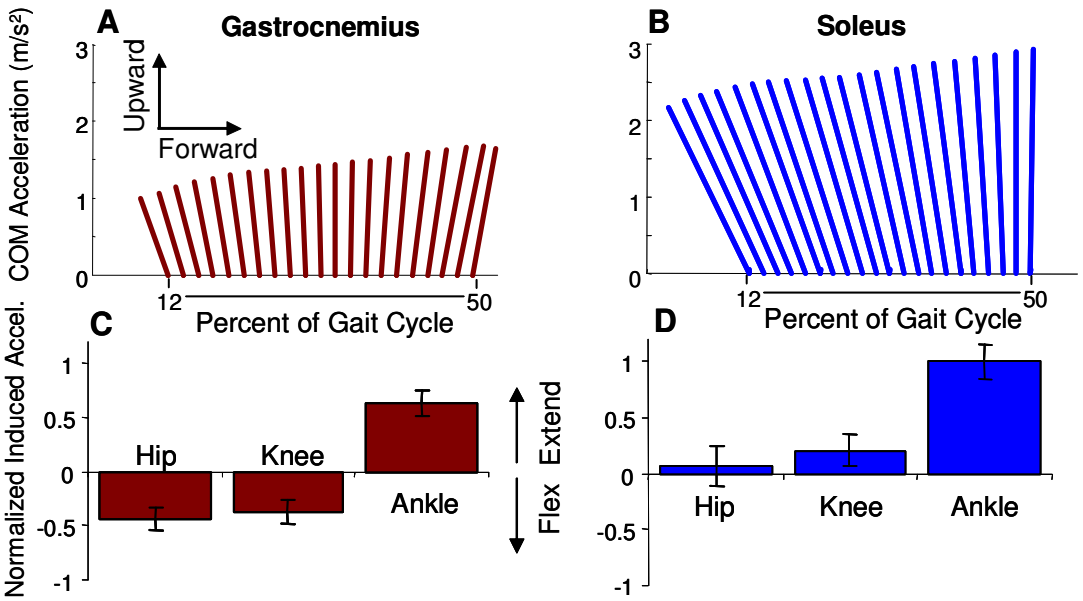

Figure 2 A,B: Average induced acceleration of the COM in the upward and forward/backward directions at each 5% of single limb stance. **C,D:** Average induced angular acceleration of hip, knee, and ankle normalized by the maximum contribution by a muscle at each joint (see methods).

Results

In all subject simulations, the gastrocnemius and soleus were highly active during single-limb stance and made large contributions to joint and COM accelerations. Throughout single limb stance, the gastrocnemius provided upward acceleration of the mass center (Figure 2A). The COM accelerations induced by the gastrocnemius were directed backward in early stance and forward in later stance. The soleus provided upward and backward acceleration of the COM in all subjects (Figure 2B). Although the gastrocnemius generated flexion accelerations of both the hip and the knee, it also provided a large plantarflexor acceleration of the ankle (Figure 2C). The soleus generated moderate extension accelerations of the hip and knee, and a strong plantarflexion acceleration at the ankle (Figure 2D).

Discussion

Dynamic simulations of subjects with crouch gait revealed that the role of soleus was to extend all the joints and to support the body's weight, consistent with findings from unimpaired individuals. Gastrocnemius also supported the body weight, but it more significantly contributed to knee and hip flexion compared to unimpaired individuals. These findings present a potential trade-off when considering lengthening gastrocnemius as a form of treatment for crouch gait in these subjects. It is important to keep in mind that all 4 subjects in this study achieved foot flat during single limb stance and that muscle contributions may differ for individuals who walk in an equinus gait. The results of this study emphasize the utility of dynamic simulations for quantitatively examining the complex actions of muscles and provide important insight into the contrasting roles of the gastrocnemius and soleus during crouch gait.

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UNDERSTANDING STIFF-KNEE GAIT - A DYNAMIC WALKING APPROACH

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Introduction

Stiff-knee gait is common in children with cerebral palsy (CP), and characterized by reduced knee flexion and knee flexion velocity in (pre)swing. It can lead to reduced clearance, frequent tripping, and reduced step length and speed, and thereby limit functional performance. Possible contributing factors include excessive rectus femoris activity in (pre)swing [1], as well as altered or diminished push-off [2,3]. In CP, stiff knee gait is often accompanied by other gait deviations, such as equinus, crouch or hyper-extended gait. These deviations lead to abnormal positioning of the leg during push-off (Fig.1), which may influence the effectiveness of push-off, for example by affecting the distribution of energy between raising the COM and progression of the leg into swing. However, there is still a limited understanding of the biomechanical factors that lead to adequate knee flexion in swing. Simple models may help improve our understanding of the dynamics of walking, and give insight into the basic concepts of push-off and swing leg characteristics in normal and stiff-knee gait.

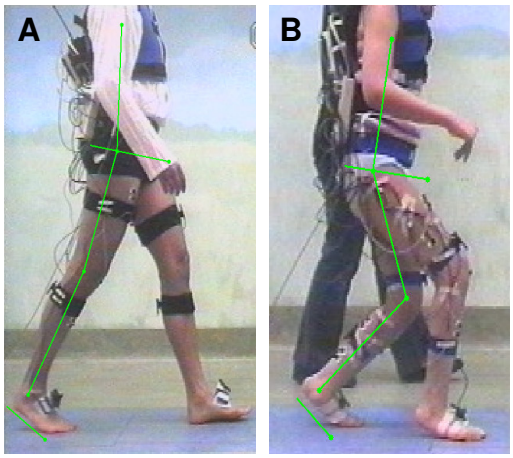

Fig.1: Knee angle at start push-off in A) normal and B) crouch gait differ vastly, which may affect the progression of the leg into swing

The **aims** of this study were: 1) to develop a dynamic walking model to study clinically relevant questions; 2) to study the intrinsic effect of a crouched posture on knee flexion in swing.

Clinical Significance A better understanding of underlying causes of stiff knee gait is essential to develop adequate treatment options.

Methods

We developed a forward dynamic simulation model of human gait, based on the simplest walking model [4-6]. The model consisted of a point mass upper body, two upper legs, two lower legs and two arc-shaped feet (Fig.2). The leg segments had anthropomorphic mass and inertia, and the point mass upper body had anthropomorphic mass. The ankles of both legs were locked at 0 degrees, so that shank and foot formed one rigid body. The knee of the stance leg was locked so that the stance leg acted as one inverted pendulum. The angle at which the stance knee was locked was varied as independent variable. The knee of the swing

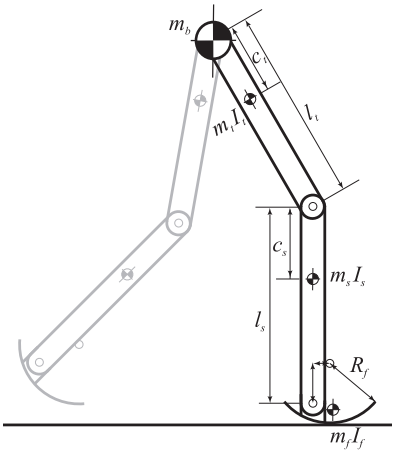

Fig.2: Representation of the model

leg was passive during swing, until it reached its preset stance leg knee angle, at which point it was locked. The hip joint was fully passive. The model was powered by a pre-emptive, instantaneous push-off impulse applied under the stance foot just before contralateral heel strike. Push-off impulse size and direction were held constant for all stance leg knee angles, and the effect of choosing different impulse size and direction values on the results was evaluated. Immediately after push-off, heel strike of the leading leg with the floor was modeled as an instantaneous, perfectly inelastic collision. Equations of motion were derived as described in [7]. A limit cycle analysis was performed to find periodic gait and stability was assessed using Floquet analysis [4]. All simulations were performed in MatLab®.

Results

With increased crouched posture, peak knee flexion in swing decreased strongly (Fig.3), resulting in a 'stiff-knee' gait pattern. Stable walking patterns could be achieved for a large range of stance leg knee angles and push-off impulse size and direction. Knee flexion in swing increased with stronger push-off impulse (Fig.4) and decreased when the impulse was directed more forward, but the difference between straight legged and crouched walking remained (Fig.4). Walking speed ranged from 0.4 to 1.2 m/s depending on push-off size and direction, and changed only slightly with crouched posture.

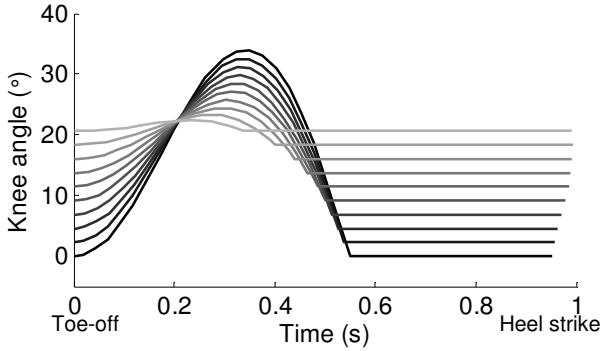

Fig.3: Knee flexion in swing as a function of time for a range of stance leg knee angles

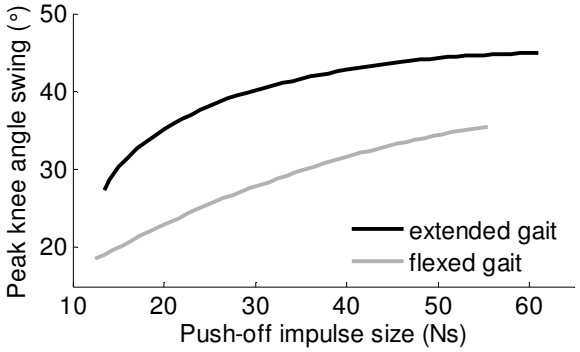

Fig.4: Peak knee flexion in swing versus push-off impulse, for extended (0º) and flexed-knee (18º) gait

Discussion

Since no confounding factors were present in the model, the decrease in swing knee flexion with crouched posture can be explained fully by the passive dynamics of the swing leg due to differences in position of the leg at swing initiation. Although the presented model clearly does not cover all characteristics of human walking, it allows studying factors that contribute to adequate progression of the leg into swing. Increasing the complexity of the model, such as replacing the push-off impulse by ankle or hip moments and finite-time double support, may provide additional insight into the dynamics and the role of push-off on swing leg knee flexion. Our findings demonstrate that decreased knee flexion in swing can occur without any differences in actuation. This indicates that part of a stiff knee gait pattern may be explained by uncontrolled dynamics of the system, rather than muscular or neurological control.

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ELECTRICAL STIMULATION OF THE RECTUS FEMORIS DURING PRE-SWING DIMINISHES KNEE FLEXION DURING SWING

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INTRODUCTION

Computer simulations have been used to predict *dynamic muscle function* during walking; i.e., how a muscle contributes to joint motions[1-3]. However, simulation predictions must be validated before they can be used as treatment guidelines for gait disorders. Electrical stimulation protocols have recently been introduced to test simulation predictions under controlled conditions[4-6]. In this study, we synchronized electrical stimulation to the gait cycle to empirically measure rectus femoris (RF) muscle function during walking. We tested the hypothesis that RF induces substantially greater knee (and hip) extension during swing if stimulated prior to toe-off rather than after toe-off.

CLINICAL SIGNIFICANCE

Stroke and cerebral palsy patients often display diminished knee flexion during swing (stiff-knee gait) and increased RF activity during pre- and early swing. The results of this study support the emerging idea that it is the RF activity during pre-swing that has the greatest potential to induce a stiff-knee gait pattern[2,7].

METHODS

Four healthy young adults (1.75±0.06m, 69±7.7kg) participated. Subjects performed 60s walking trials on a treadmill at preferred speed. A real-time controller delivered short-duration electrical stimuli at either 50% (pre-swing) or 60% (early swing) of the gait cycle, in approximately 6 randomly selected strides per trial. An individual stimulus consisted of four 300μs current (≤50mA) pulses delivered over 90 ms to the rectus femoris via surface electrodes. EMG signals were recorded and used to determine the onset of stimulations, and to evaluate potential stimulation to neighboring muscles. Whole body kinematics were simultaneously recorded and used to compute lower extremity joint angles. The stimulation-induced movement was defined as the deviations of the hip and knee flexion angles from expected values at the point of peak knee flexion during swing. A forward dynamic simulation of normal walking was used to simulate the experimental conditions[8]. Hip and knee angle deviations were compared between non-stimulated trials, stimulated trials, and the simulations.

RESULTS

Knee and hip flexion reductions averaged 0° for all non-stimulated trials (Fig. 1). Decreases in knee flexion averaged 10° across subjects when the stimulus was introduced during pre-swing, compared to 1° when the stimulus was introduced during early swing. We also measured small decreases in average hip flexion for the pre-swing (3°) and early-swing (1°) stimulations. The measurements were generally consistent with our model predictions that knee and hip flexion would decrease by 12° and 2°, respectively, for stimulation applied during pre-swing, and by 6° and 0° for stimulation applied during early swing.

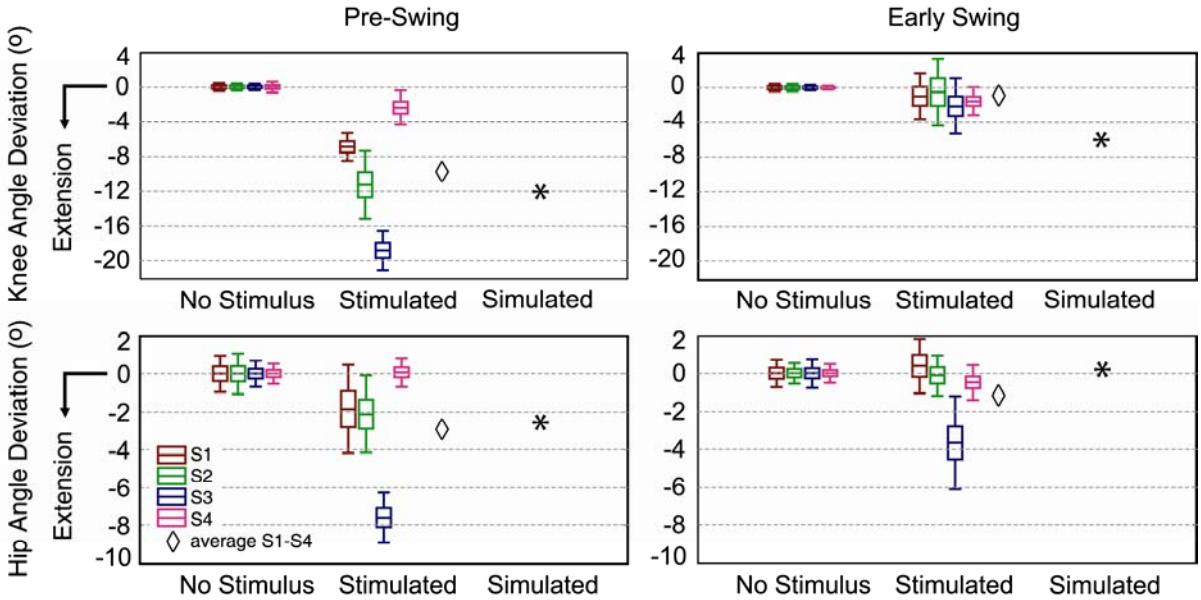

Figure 1: Hip and knee flexion deviations at the point of peak knee flexion during swing for non-stimulated and stimulated trials during pre-swing (left) and early swing (right). S1, S2, S3 and S4 = subjects 1 through 4. ◊ = average of S1-S4 for stimulated trials. Simulation predictions (*) have been included for comparison

DISCUSSION

Our experimental results confirm the simulation-based hypothesis that RF has a substantial effect on peak knee flexion during swing if stimulated prior to, but not after toe-off. Hence, stiff-knee gait patterns may partially arise from abnormal RF activity during pre-swing[2,7]. However, there was substantial variability in the magnitude and type of motion induced by RF stimulation in our subjects. Hence, it is important to ascertain what factors (e.g., musculoskeletal geometry, walking kinematics, or coordination patterns) contribute to this variability, particularly when using computer simulations to guide pathological gait treatment. Future modeling efforts should assess the influence of variability in model parameters on dynamic muscle function.

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Mechanism of improved knee flexion after rectus femoris transfer surgery

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Introduction

One of the most common gait problems in children with cerebral palsy is the inability to appropriately flex the knee during the swing phase of walking, or "stiff-knee gait" [1,2]. Rectus femoris transfer surgery is frequently performed to treat stiff-knee gait in subjects with cerebral palsy. In this surgery, the distal tendon is released from the patella and re-attached to one of several sites, such as the sartorius or the iliotibial band. Surgical outcomes vary [3,4], and the mechanisms by which the surgery improves knee flexion are unclear. It has been suggested that transferring the rectus femoris converts the muscle from a knee extensor to a knee flexor, thereby increasing knee flexion [5]. However, experimental studies have found that the muscle produces a knee extension moment after surgery [6,7], possibly due to scarring to underlying tissue [8]. The purpose of this study was to examine the mechanism by which the rectus femoris transfer surgery increases knee flexion.

Clinical Significance

It is difficult to improve surgical outcomes because the mechanism by which knee flexion increases in some patients is unknown. This study examines three types of transfer to clarify the mechanism by which the transferred muscle improves knee flexion.

Methods

Muscle-driven simulations were created of ten children diagnosed with cerebral palsy and stiff-knee gait. These simulations were altered to represent surgical transfers of the rectus femoris to the sartorius (Fig. 1b) and the iliotibial band (Fig. 1c). Rectus femoris transfers in which the muscle remained partially attached to the underlying vasti through scar tissue were also simulated by reducing the muscle's knee extension moment (Fig. 1d).

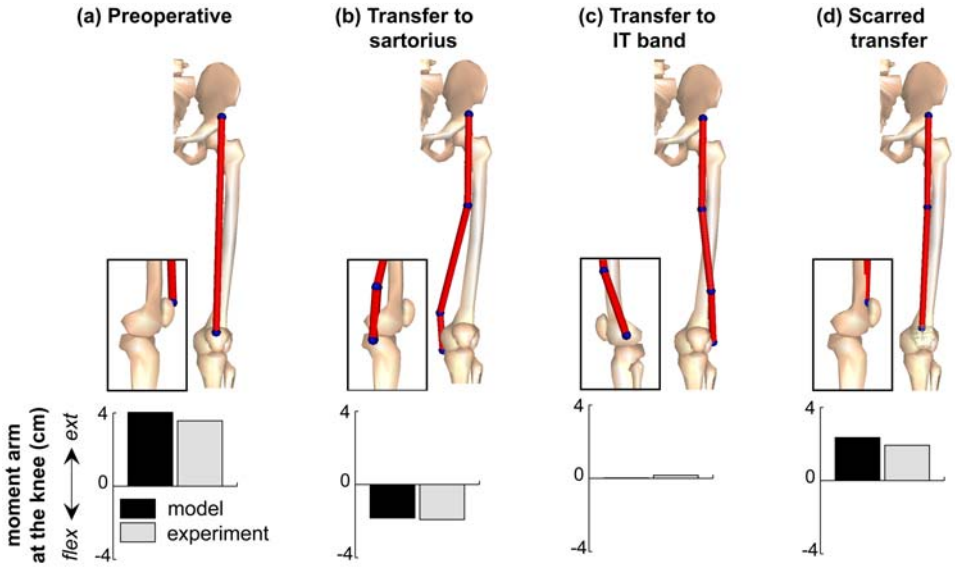

Figure 1. Illustration of rectus femoris in each musculoskeletal model with plots of moment arms at the knee (averaged over 20 - 60 degrees of knee flexion) compared to experimental data [7,9].

Results

Simulated transfer to the sartorius, which completely converted the rectus femoris' knee extension moment to a flexion moment, produced 32° ± 8° improvement in peak knee flexion on average (Fig. 2). Simulated transfer to the iliotibial band, which eliminated the muscle's knee extension moment but did not convert it to a flexion moment, predicted only slightly less improvement in peak knee flexion (28° ± 8°). Scarred transfer simulations, which reduced the muscle's knee extension moment, predicted significantly less ($p < 0.001$, paired t-test) improvement in peak knee flexion (14° ± 5°).

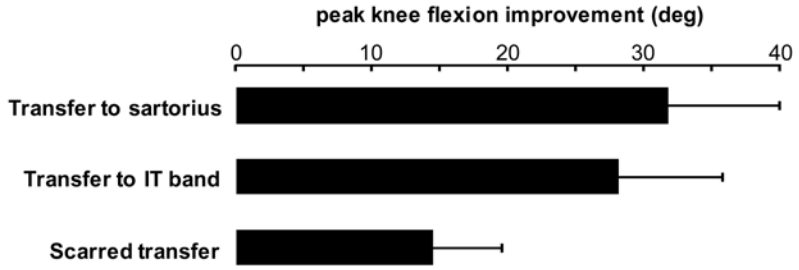

Figure 2. Increases (plus one standard deviation) in peak knee flexion averages over the ten subjects for each of the simulated treatments.

Discussion

Simulated transfer to the sartorius, which converted the rectus femoris' knee extension moment to a flexion moment, predicted only 4° greater knee flexion improvement on average than transfer to the iliotibial band, which eliminated the muscle's knee extension moment. Significant improvement in knee flexion was attained even when the extension moment of the rectus femoris was reduced. These findings suggest that the primary mechanism for improvement in knee flexion after surgery is reduction of the muscle's knee extension moment, rather than conversion to a knee flexion moment. Scarred transfer simulations resulted in an average peak knee flexion improvement (14° ± 5°) comparable to the average increase in knee flexion range of motion (13° ± 11°) in the subjects' postoperative data. Simulated non-scarred transfers to the iliotibial band and the sartorius predicted greater improvements in knee flexion than the scarred transfer simulations, suggesting methods to reduce scarring may further augment knee flexion.

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Long Term Follow-Up After Comprehensive Clubfoot Surgery

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Clinical Significance: Long-term follow-up is crucial for evaluating the efficacy of clubfoot treatment. Little is known about the function of the surgically treated clubfoot or patient satisfaction as young adults enter the work force. The study objective was to quantify the long-term outcomes of adult subjects who underwent surgical releases for clubfoot as infants.

Introduction: Idiopathic clubfoot (talipes equinovarus) is one of the most common congenital deformities. It is characterized by equinus contracture at the ankle and inversion of the hindfoot, and midfoot through the axis of the subtalar joint. The goal of treatment is a pain-free, functional, plantigrade foot which promotes a normal gait pattern. The long-term effects of surgical treatment are still unclear, but some investigators have reported residual foot and ankle weakness, loss of range of motion (ROM), and deformity which may cause impairment later in life.[1,2]

Methods: Sixteen subjects ("Clubfoot Group"; 13M, 3F, mean age 21±1.6 years) participated in this IRB-approved study. All subjects underwent comprehensive soft-tissue release for clubfoot deformity by the same surgeon between 1982–1987 (mean age at surgery 8 months; mean follow-up 21 years). Twenty-one age matched controls ("Control Group"; 12M, 9F, 21.9±2.9 years) were evaluated for comparison. Gait temporal and spatial parameters were measured and analysis of foot kinematics were performed using a multisegmental foot and ankle model.[3] Isokinetic testing was performed to assess ankle plantar flexion, inversion, and eversion muscle strength using the Biodex System 3 (Shirley, New York). Subjects also completed validated outcomes instruments including the Short-Form 36 (SF-36) and the midfoot and hindfoot portions of the American Orthopaedic Foot and Ankle Society (AOFAS) outcomes tool.

Results: Statistical differences in kinematics of the hindfoot and forefoot were noted between the Clubfoot Group and Control Group (Figure 1). The Clubfoot Group displayed reduced hindfoot ROM in the sagittal and coronal plane and decreased hindfoot dorsiflexion throughout the stride. The Clubfoot Group also demonstrated significantly less forefoot plantarflexion compared to the Control Group throughout the gait cycle. Diminished stride length, walking speed, and increased stance phase duration were also observed (Table 1). The Clubfoot group also exhibited reduced strength in all tested ankle motions, significantly lower hindfoot and midfoot AOFAS scores, and significantly lower values for the SF-36 bodily pain subscale.

Discussion: The results of this study demonstrate measurable differences between the Control Group and Clubfoot Group. The Clubfoot Group exhibited significant deviations in the forefoot position and limitations in hindfoot motion. Structural abnormalities and pain may be associated with decreased strength during ankle motions such as plantar/dorsiflexion and inv/eversion. This relationship is supported by the reduced temporal spatial parameters, lower AOFAS scores and increased bodily pain.

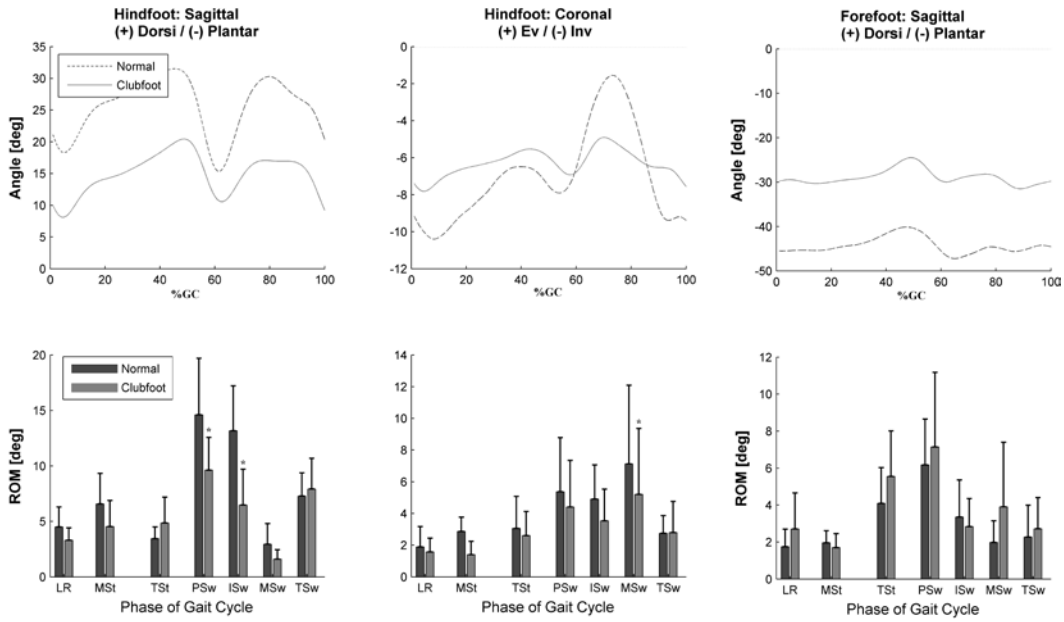

Figure 1. Intersegmental kinematics and ROM of the hindfoot and forefoot. The asterisk (*) represents significant differences between groups ($p<0.05$).

Table 1.

Temporal Spatial Parameters (St. Dev)				
	Walking Speed (m/s)	Cadence(strides/min)	Stride Length(m)	Foot Off (%GC)
Clubfoot Group	0.99(0.14)*	104.1(9.0)*	1.14(0.1)*	62.6(1.9)*
Control Group	1.22(.13)	110.6(6.8)	1.32(0.1)	60.3(1.0)
Peak Torques N-m (St. Dev) Normalized to Body Weight				
	Plantarflexion	Dorsiflexion	Eversion	Inversion
Clubfoot Group	29.9(28)*	23.4(17)*	14.4(11)*	15.4(10)*
Control Group	92.6(33)	43.0(13)	24.2(7)	25.9(9)
AOFAS Scale Scores (St. Dev)				
	Hindfoot	Midfoot		
Clubfoot Group	79.6(10)*	79.7(11)*		
Control Group	99.7(1.7)	99.8(1.3)		
SF-36 Scores				
	Physical Function	Bodily Pain	General Health	Social Functioning
Clubfoot Group	86.8(14)*	61.3(27) *	78.4(14) *	94.5(9)
Controls Group	99.0(3.9)	91.9(11.1)	87.5(12.4)	97.6(6.7)

*Indicates significantly different from Control Group ($p<0.05$)

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Acknowledgements

Conservative and Surgical Clubfoot Treatment: Multi-Center Study. U.S. Department of Education. **NIDRR Grant H133G060252**

A Comparison of Children with a History of Clubfoot Which Underwent Surgical or Ponseti Treatment

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Introduction

Idiopathic clubfoot is the most common congenital deformity of the foot[1]. Most orthopedic surgeons agree that initial treatment should be non-operative, but there is controversy over whether operative intervention is more or less beneficial than continued non-operative treatment[1]. Previous studies have looked at effectiveness of treatment in children with clubfeet[2], but none have included a comprehensive comparison including instrumented gait analysis with multi-segment foot kinematics, foot pressure, and physical exam.

Statement of Clinical Significance

The purpose of this study is to complete a comprehensive comparison of outcomes in patients with a history of clubfoot treated surgically or with the Ponseti technique.

Methods

Children in two groups were evaluated retrospectively: clubfoot, treated operatively (n=26, feet= 43, age 5-11), clubfoot, treated with Ponseti technique (n=22, feet=35, age 5-10). Children in the operatively treated group had traditional posterior medial releases following initial casting, with 10 patients requiring repeat surgical intervention. The Ponseti group was treated with an average of 5.0 casts, and 18 of 22 had Achilles tenotomies. Five patients in the Ponseti group relapsed requiring additional casting and in some cases surgery (2 patients anterior tibialis tendon transfer, 1 Achilles lengthening). Foot motion was measured utilizing a multi-segment foot marker set in the operative group (figure 1).[3] Results were compared to normative data collected from 34 children (4 to 17 years old) with no known foot deformity. Reflective markers were placed over boney prominences and recorded during walking using an eight camera Motion Analysis system (Motion Analysis Corp., Santa Rosa, CA). This multi-segment foot model allowed the calculation of the position and orientation of the hindfoot, medial forefoot, lateral forefoot, and the hallux. The medial longitudinal arch of the foot was characterized by the ratio of its height to foot length (arch index). The Ponseti group was evaluated using a traditional single segment foot model with Helen Hayes marker set. Physical exam (PE) measurements were recorded, and dynamic foot pressure readings were taken (Tekscan). Statistical significance was determined using t-tests.

Figure 1: Multi-segment foot model markers.

Results: Table 1: Normal/Ponseti/Operative group treatment results.

Result	Normal	Clubfoot Ponseti	Clubfoot Operative
Multi-segment Foot Kinematics			
Hindfoot DF/PF range	21.8±3.4	NA	15.8±3.3*
Hindfoot valgus mean stance	1.2±4.7	NA	0.2±5.2
Arch height	0.26±0.04	NA	0.27±0.06
Forefoot abduction mean stance	11.4±4.5	NA	10.6±7.7
Single Segment foot Kinematics			
Ankle DF/PF range	28.3±4.1	24.2±2.8*#	19.4±3.3*
Ankle max DF stance	13.9±2.3	12.4±3.1*	11.6±6.3
Ankle push off power generation	24.9±6.5	18.2±5.8*#	13.5±3.9*
Foot Pressure			
Varus (-)/valgus(+) score	11.0±23.6	-15.7±18.9*#	-36.8±24.7*
Heel impulse	37.3±9.6	42.2±16.0#	32.2±14.4
Medial forefoot pressure	33.5±9.2	39.8±11.9*#	19.1±8.8*
Lateral midfoot pressure	7.1±6.3	22.9±7.7*	25.2±10.8*
Physical Exam			
DF with knee extended PROM	8.4±7.1	9.7±5.2#	-0.6±8.6*
PF PROM	59.4±7.9	51.3±10.3*#	28.0±10.7*
DF/PF passive range	67.8±10.5	61.0±12.6*#	27.5±9.2*
Thigh foot angle	7.8±5.9	13.6±8.1*#	2.2±13.9*
Forefoot adduction	NA	6.2±5.8#	10.1±7.6
Calcaneal inversion/eversion PROM	NA	37.6±12.4#	17.1±8.7

*Different from normal ($p<0.01$) # Different from operative ($p<0.01$) NA=not available

Discussion: Physical exam results demonstrate severe reduction of motion in operatively treated feet, and only mild reduction of motion in patients treated with the Ponseti method. Foot pressure results in the operatively treated group show residual varus with reduced medial forefoot pressure and elevated lateral midfoot pressure. The Ponseti group showed mild residual varus. While sagittal plane ankle PROM from PE is severely reduced in the operatively treated group, only mild limitations are noted in the sagittal plane kinematics. The Ponseti group is again only mildly limited. Power generation reveals significant limitation in the operative group, while the Ponseti group is only mildly limited. Multi-segment foot kinematics from the operative group demonstrates overall good alignment, but reduced motion compared to normative data. While position is relatively well corrected in the operatively treated group, motion and power generation continue to be limited. In general, the average operatively treated patient seemed to have sufficient sagittal plane motion for walking. With motion limitations compared to normal and Ponseti treated feet, greater functional limitations may be noted in the operatively treated group in higher-level skills such as running and jumping. Additional study is needed to look at multi-segment kinematics in the Ponseti treatment group.

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ANKLE KINEMATICS AND KINETICS FOLLOWING NON OPERATIVE TREATMENT FOR CLUBFOOT

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Introduction

Previous research has shown clubfoot patients to have weak, stiff ankles following surgical intervention[1] leading our institution to look at the efficacy of non operative treatments including Ponseti casting (Ponseti) and French physiotherapy (PT) as alternatives to surgical management.[2]

Preliminary kinematic results at age two years, showed 47% of Ponseti feet and 65% of PT feet to have normal sagittal ankle motion.[3] Outliers were identified, showing a small number of PT feet with residual equinus while almost half of Ponseti feet had excessive dorsiflexion in mid stance. It is unknown if these findings are a result of the percutaneous tenotomy commonly performed in the Ponseti method, while absent in the PT protocol. These findings were reported in very young children, and it is unknown if these residual deformities resolve with time, or if they require surgical management in the future.

This study was designed as an intermediate follow up to determine if ankle motion and push-off power were better following non operative treatment, than in patients treated surgically. Data were collected on three patient groups: 1) *Ponseti*, 2) *PT* and *3) Surgical* (initially treated non operatively). All groups were compared to a group of age matched controls.

Clinical Significance

Long term prospective studies have helped clinicians objectively assess the ankle function following non operative treatment protocols. Since the non operative approach for treating clubfoot has gained the attention of the orthopedic community, it is imperative to study the outcomes following these procedures and compare them to both patients who have undergone surgery and to normal controls.

Methods

Clubfoot patients treated non operatively underwent gait analysis at approximately five years of age, as part of a prospective study. Patients walked at a self selected speed while kinematic and kinetic data were collected using a VICON 512 motion capture system (Vicon, Oxford, England), and two ATMI force plates (Watertown, MA). Ankle data were reduced using Vicon Clinical Manager (Vicon, Oxford, England) and a representative trial was used for analysis. The following variables were identified: maximum dorsiflexion in 2nd rocker, maximum plantarflexion in 3rd rocker, maximum dorsiflexion and maximum plantarflexion over the gait cycle were added to assess total dynamic range of motion (ROM), and the peak ankle moment and power. Data points were averaged and presented as a mean with the corresponding standard deviation (in parentheses). Between group comparisons were made with an ANOVA ($p<0.05$) and a post hoc Tukey test was run to determine level of significance. A student T-test was used to determine difference in average power in patients who had a tenotomy and those who did not. Seventeen age matched control subjects were used for comparison.

Results

Ninety clubfoot patients were tested on average at age 5.25 years (±0.33). At the time of testing, 34 *Ponseti* feet and 40 *PT* feet remained non operative and there were 51 *Surgical* feet. Kinetic data was collected on all but seven limbs (due to stride length or cooperation). Kinematic and kinetic results can be found in Table 1.

Maximum plantarflexion in 3rd rocker was significantly less in the *Surgical* group compared to *Control* (p=0.0203). Both total dynamic ROM and peak ankle power were significantly less in both the *Surgical* and *PT* groups than in *Controls* (p=0.0044 and p=0.0003, respectively). The average peak ankle moment was less in *Ponseti, PT* and *Surgical* compared to *Control* (p<0.0001). No difference in ankle power was found (p=0.222) in the non operative feet who had a tenotomy (n=28) verses those who did not (n=42).

		Kinematics				**Kinetics**	
Group	# Feet	Max DF° (2nd Rocker)	Max PF ° (3rd Rocker)	Total Dynamic ROM °	# Feet	Ankle Moment (Nm/kg)	Ankle Power (W/kg)
Ponseti	34	12.5 (±4.2)	15.4 (±6.7)	30.1 (±5.9)	32	0.74 (±0.11)	2.36 (±0.71)
PT	40	10.4 (±4.0)	16.2 (±8.0)	28.2 (±6.7)	38	0.76 (±0.14)	2.21 (±0.83)
Surgical	51	12.5 (±5.2)	12.7 (±5.9)	27.2 (±4.7)	48	0.75 (±0.13)	1.97 (±0.62)
Control	17	12.1 (±4.3)	**18.1**† (±8.0)	**32.7**[3] (±6.5)	17	**0.95**‡ (±0.12)	**2.83**[3] (±0.64)

Table 1. p<0.05: †Normal > Surgical; [3]Normal > Surgical/PT; ‡Normal > Ponseti/PT/Surgical

Discussion

No differences were found in kinematic or kinetic variables between clubfoot treatment groups, be it non operative or operative. It is clear that the *Surgical* feet lacked plantarflexion at push off and over all Dynamic ROM, leading to a decreased moment and a lack of ankle power generation at push off. The *PT* feet also showed decreased dynamic ROM, ultimately leading to poor ankle power. The *Ponseti* feet had normal ankle motion and although the peak moment was significantly less than normal, ankle power was not different from normal.

Results following non operative clubfoot treatment are encouraging; however results did not show clear distinction between non operative and operative feet. Significant differences from normal were found in total dynamic range of motion, but the clinical significance of these findings may be questionable, when the difference in means is as small as 4.5 degrees (*PT* group).

Early kinematic findings on toddlers suggested *Ponseti* feet may have predicted poor push off power due to the percutaneous tenotomy, but follow up at five years of age, does not support these findings. Further work needs to be done looking at motion patterns in feet that are clear outliers (ie: residual equinus, excessive dorsiflexion, and calcaneus) and the corresponding kinetic data to best study the treatment effect in feet that were corrected non operatively to those that had surgical release.

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Plantar Pressures Vary Across Elliptical Trainers and Compared to Walking

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INTRODUCTION

Following a diagnosis of Type 2 Diabetes, many individuals are instructed to engage in cardiovascular exercise to help manage blood glucose levels while also maintaining cardiorespiratory fitness, body composition, strength and endurance. The American College of Sports Medicine recommends performing 10 to 30 minute sessions, up to five times a week.[1] One form of cardiorespiratory exercise, elliptical training, is gaining popularity not only in fitness facilities and homes, but also in rehabilitation settings. This exercise has notable movement similarities to walking.[2,3] However, it lacks periods of single limb support as both limbs maintain contact with the support surface throughout the movement cycle. One area that has not been systematically studied is what impact the continuous double limb support has on pressure patterns experienced by the feet. We hypothesized that, when compared to walking, peak pressures would decrease while elliptical training due to the shared load between the supporting limbs. We expected the reduction in peak load to offset the extended support time, resulting in no significant differences in pressure-time integrals between walking and elliptical training.

CLINICAL SIGNIFICANCE

Prolonged and repetitive exposure to elevated plantar pressures can lead to tissue injury and foot ulcers in individuals with neuropathic foot disorders arising from Type 2 diabetes. Improved understanding of how plantar pressures vary across elliptical trainers will help guide clinicians, exercise physiologists and individuals with diabetes in the selection of exercises that promote therapeutic and fitness goals while minimizing the risk of foot injury.

METHODS

Twenty individuals (10 males, 10 females; mean age 46 years) without known neurologic or orthopedic injuries participated. Following two training sessions on the elliptical trainers (Life Fitness, SportsArt, Octane, True), plantar pressure variables (Pedar) and support surface kinematics (Qualisys) were recorded as subjects walked and elliptical trained at a self-selected speed. Elliptical training order was randomized. Data were recorded during the third minute of each exercise. Peak pressures and pressure-time integrals were identified for the heel, arch, and forefoot. The influence of activity on pressure variables in these anatomic regions was determined using separate 5x1 analyses of variance with repeated measures for activity. When assumptions of normality were violated, the Friedman Repeated Measures ANOVA on Ranks was performed.

RESULTS

Peak Pressure (Figure 1): *Heel* peak pressures were significantly higher during walking (189 kPa) compared to the True (107 kPa), Octane (121 kPa) and Life Fitness (125 kPa) conditions ($p<0.05$), but not the SportsArt (144 KPa). Heel pressures on the SportsArt exceeded those on the True ($p<0.05$). Peak pressure under the *Arch* varied significantly only between the walking (113 kPa) and True (90 kPA) conditions ($p<0.05$). *Forefoot* peak

pressures were significantly higher while walking (232 kPa) compared to the True (157 kPa), Life Fitness (153 kPa), Octane (144 kPa) and SportsArt (139 kPa) conditions ($p<0.05$), but did not vary significantly among elliptical trainers.

Pressure-Time Integral (Figure 2): *Heel* pressure-time integrals on the SportsArt [73 (kPa)*s], Octane [50 (kPa)*s], and during walking [50 (kPa)*s] were significantly longer than the True [38 (kPa)*s; $p<0.05$]. Pressure-time integrals on the Sports Art also exceeded those on the Life Fitness [45 (kPa)*s; $p<0.05$]. Pressure-time integrals under the *arch* differed significantly only between the SportsArt [48 (kPa)*s] and True [41 (kPa)*s; $p<0.05$]. *Forefoot* pressure-time integrals did not differ significantly across conditions.

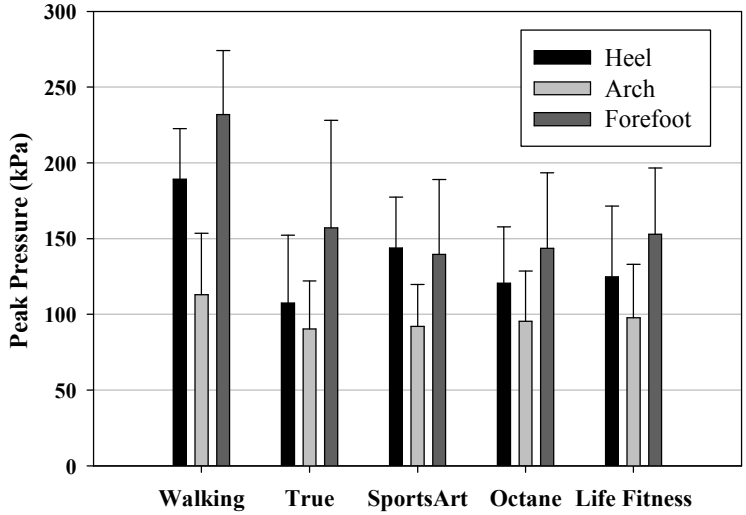

Figure 1. Peak pressure values under heel, arch and forefoot during walking and elliptical training (mean and standard deviation; N=20).

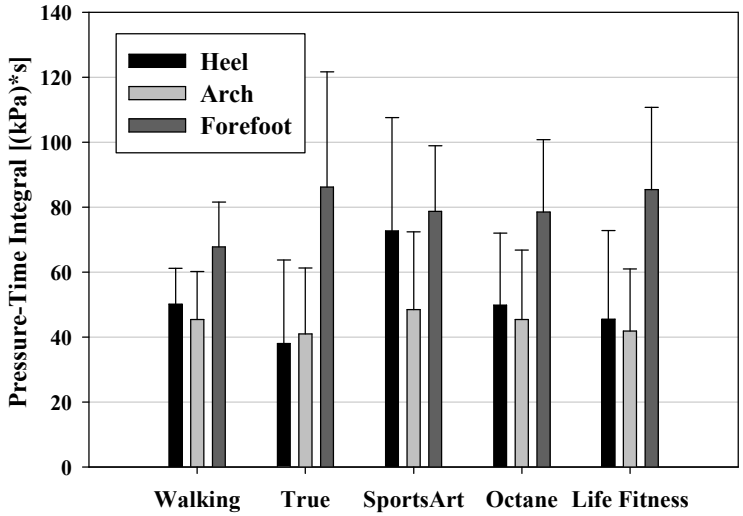

Figure 2. Pressure-time integrals under heel, arch and forefoot during walking and elliptical training (mean and standard deviation; N=20).

DISCUSSION

The forefoot is one of the most frequent locations for diabetic foot ulcers. Peak pressures in this region were consistently lower during elliptical conditions compared to walking. Forefoot pressure-time integrals did not vary significantly between walking and elliptical training, however high inter-subject variability may have masked differences. Additional work is underway focusing on the influence of elliptical training with varying stride lengths (longest vs. shortest) on plantar pressure variables to further refine treatment recommendations for use of this therapeutic tool.

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Correlation between plantar pressure and Oxford Foot Model kinematics in clubfoot

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INTRODUCTION

Plantar pressure measurements and multi-segment foot model kinematics are used in clinical gait analysis to assist in the objective measurement of dynamic foot deformity. The aim of this study was to correlate the findings of pressure measurements with Oxford Foot Model (OFM) kinematics [1] to provide a comprehensive assessment of foot deformity.

CLINICAL SIGNIFICANCE

Correlating multi-dimensional foot kinematics with plantar pressure measurements enables advanced understanding of dynamic foot deformity and its effects on forces under the foot. It provides information on primary and secondary deformities and guides clinical management.

METHODS

11 children with idiopathic clubfoot treated at a young age with a postero-medial release were assessed (6 male, 5 female, age 10.5±3.3 yrs) using lower limb kinematics, Oxford Foot Model (OFM) [1] kinematics and plantar pressure. Kinematic data was collected with a 12 camera Vicon 612 system (Oxford, UK). Plantar pressure data were collected with a prototype, piezo-resistive pressure plate (Istituto Superiore di Sanita, Rome, Italy) with a spatial resolution of 5 mm [2]. The OFM markers were superimposed onto the pressure footprint during mid-stance. The co-ordinates of each marker were then projected vertically onto the footprint (Figure 1). This allowed the foot to be divided into five sub-sections on the basis of anatomical landmarks, and to correlate pressure findings with the output from the OFM. Peak force from each subdivision was correlated with clinically relevant variables from the OFM. For the purpose of this study, only the affected limb was analyzed; for children with bilateral clubfoot, the right leg was analyzed.

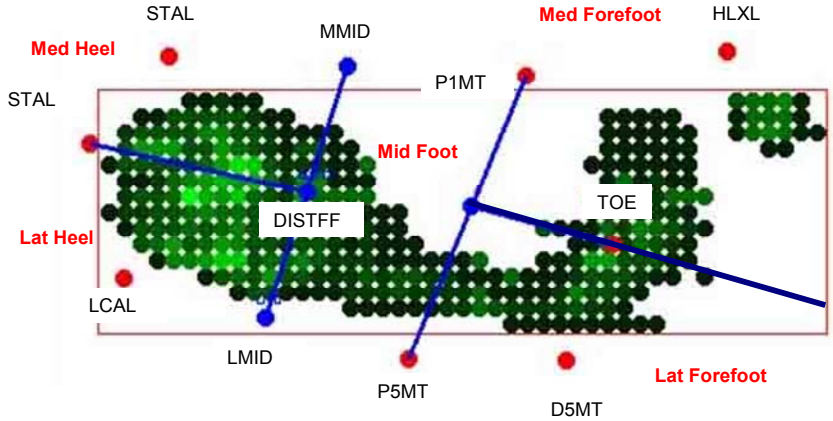

Figure 1: Pressure footprint showing five sub-areas. The labeled circles represent the projected positions of markers on the foot.

RESULTS

A significant inverse correlation was found between forefoot supination (in relation to the hindfoot) and the medial/lateral distribution of force at the hindfoot (Table 1). An inverse

correlation was also found between increased forefoot dorsiflexion (in relation to the tibia) and reduced forefoot force as expected. Using independent t-tests ($p<0.05$) compared to healthy controls, the force in the midfoot ($p<0.00$) and lateral forefoot ($p=0.04$) were increased. The force in the medial forefoot was reduced ($p<0.00$). Comparing the OFM kinematics to healthy controls showed a significant reduction in dorsiflexion at the hindfoot ($p=0.01$), at the forefoot ($p<0.00$), and of the forefoot in relation to the tibia ($p<0.00$). Maximum dorsiflexion in swing was also significant at the forefoot ($p=0.01$) and of the forefoot in relation to the tibia ($p=0.03$).

Foot Model	Pressure Plate	Correlation
HF Varus	Lat:Med heel force	0.35
HF Varus	Lat heel force	0.07
HF Varus	Midfoot force	0.26
FF supination	Lat:Med FF force	-0.07
FF/Tibia supination	Lat:Med FF force	0.09
FF supination	Lat:Med heel force	-0.52
FF/Tibia supination	Lat:Med heel force	-0.31
HF Varus	Lateral FF force	0.11
HF Varus	Lat:Med FF force	0.10
HF dorsiflexion	Heel force	-0.22
FF dorsiflexion	FF force	-0.34
FF/Tibia dorsiflexion	FF force	-0.48

Table 1: Correlation between OFM and pressure plate results.

FF = forefoot, HF = hindfoot

DISCUSSION

The OFM kinematics showed a trend towards increased hindfoot varus and forefoot supination and the plantar pressure data showed increased lateral hindfoot force and reduced medial forefoot force. However no significant correlation was found between hindfoot varus and lateral heel force or forefoot supination and lateral forefoot force. The inverse correlation found between forefoot supination (in relation to the hindfoot) and lateral/medial heel force therefore indicates that forefoot pronation compensates for hindfoot varus. The force at the midfoot was significantly higher than the healthy population regardless of whether the hindfoot was in a varus or valgus position. If the midfoot was sectioned into medial and lateral components it would likely show a varus hindfoot with increased weight bearing in the lateral midfoot, and a valgus hindfoot with increased force under the arch medially. Interestingly changes in foot shape did not seem to affect hindfoot loading when compared to healthy controls. This analysis has also been completed in children with hemiplegia [3]. Interestingly in the hemiplegia population there was a correlation between forefoot loading influencing hindfoot positions whereas in the clubfoot population the strongest correlation showed hindfoot loading influences forefoot positioning.

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Kinematic Analysis of Foot Motion in Children with Diplegic and Hemiplegic Patterns of Cerebral Palsy

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INTRODUCTION

Advanced techniques in gait analysis can be used to help quantify abnormal foot mechanics during gait in children with CP and can identify unique patterns in children with unilateral vs. bilateral motor impairments. The literature points out clinically observed differences in foot deformities[1] and differences in foot pressure between children with diplegia vs. hemiplegia.[2] More work is needed to examine differences in dynamic foot deformities between children with diplegic and hemiplegic patterns of CP.

CLINICAL SIGNIFICANCE

The purpose of this study is to identify differences in intrinsic foot motion during walking in children with hemiplegia and diplegia and to examine young vs. older age groups. Results will help clinicians gain insight into how dynamic foot deformities vary in children with unilateral vs. bilateral motor impairments and how age influences foot deformity. Such understanding will lead to better treatment planning across the lifespan for individuals with CP.

METHODS

Retrospective analysis of foot kinematics was performed on a group of 53 children with CP tested in the Gait Lab between 2006 and 2008. Exclusions included children who had orthopedic surgery to correct foot deformity and/or children who had selective dorsal rhizotomy (SDR) or a baclofen pump. Other factors, such as soft tissue lengthening, boney surgery above the foot, brace wear, and history of Botox were not controlled for in this study. Both right and left sides were included in the analysis of the diplegic group and only the involved side was included for the hemiplegic group. Foot motion was measured utilizing a multi-segment foot model in which reflective markers were placed over boney prominences and recorded during walking using an eight camera, high speed motion capture system. The model allowed for calculation of the position and orientation of the hind foot, medial forefoot, lateral forefoot, and hallux. The medial longitudinal arch was characterized by the ratio of foot height to length (arch index).[3]

RESULTS

Table 1. Descriptive Summary: Diplegia vs. Hemiplegia Groups

N=53	CP Diplegia (n=25)		CP Hemiplegia (n=28)	
groups	young (n=12)	older (n=13)	young (n=14)	older (n=14)
mean age (range)	5.4 yrs (5.0 – 5.9 yrs)	13.7 yrs (11 – 16.6 yrs)	5.6 yrs (5.1 - 6.8 yrs)	12.8 yrs (10.7 – 17.7 yrs)
gender	4 girls 9 boys	4 girls 9 boys	4 girls 10 boys	4 girls 10 boys
GMFCS	I =4, II= 4, III = 4	I =5, II= 4, III = 4	I =10, II= 4	I =10, II= 4

A number a differences in segmental foot motion were identified between children with diplegia and hemiplegia (Table 2). Examination by age in each group reveled mostly

subtle differences that were not statistically significant in the diplegic children. There were greater differences between age groups in the children with hemiplegia, a number of which were statistically significant (Table 2).

Table 2. Kinematic foot motion variables for mean stance phase values

Foot Kinematic Variable (normal)	**Diplegic CP**		**Hemiplegic CP**	
	young	older	young	older
Hindfoot Varus / Valgus (1◦ valgus)	4◦ valgus		0◦ val	
	7◦ val †	1◦ val †	4◦ val ‡	3◦ var ‡
Arch Index (0.26)	0.22*		0.26*	
	0.22	0.22	0.25	0.27
Hallux Varus \ Valgus (0.5◦ valgus)	9◦ valgus*		2◦ valgus*	
	9◦ val	9◦ val	1◦ var ‡	6◦ val ‡
DF of Medial Forefoot (5◦ DF)	6◦ DF*		0◦ DF*	
	4◦ DF	7◦ DF	6◦ DF ‡	4◦ PF ‡
AB /Add of Lateral FF (9◦ abd)	8◦ abd*		0◦ abd*	
	9 ◦ abd	7◦ abd	6◦ abd ‡	5◦ add ‡

($p<.01$)* diplegia vs. hemiplegia † (y vs. old diplegia) ‡ (y vs. old hemiplegia)

DISCUSSION

Results confirm clinically observed differences in patterns of foot deformity between children with hemiplegic vs. diplegic cerebral palsy. Children with hemiplegia assume positions of hindfoot varus, and forefoot adduction /plantarflexion. These deviations are more pronounced in an older hemiplegic group. Children with diplegia tend to assume positions of hind foot valgus, and hallux valgus. Except for less hindfoot valgus in an older group, the differences with age are not as striking. The reasons for this are not clear. One explanation could be that valgus foot deformities are static in children with diplegia; however we do not observe this clinically. Our hypothesis is that within the population of children with diplegic CP, there is a subgroup in which early valgus deformities resolve and a subgroup whose valgus deformities progresses with age. The study here is unable to explore this hypothesis due to the lack of a longitudinal design and small numbers in each group. This work does provide evidence that a multi-segment foot model is able to measure clinically important differences in the patterns of intrinsic foot deformity between children with unilateral vs. bilateral motor impairments.

We are currently following young children with CP in a long-term study and hope that results from our study will provide evidence of which factors predict how foot deformities progress or resolve as children with CP grow.

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KINEMATICS OF THE UNAFFECTED FOOT IN HEMIPLEGIC CEREBRAL PALSY

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INTRODUCTION

Initial examination of the foot model kinematics of the unaffected foot of hemiplegic cerebral palsy (CP) subjects showed a range of abnormalities [1]. It was thought that these abnormalities could either be compensations or due to mild neurological involvement of this side. Multi-segment foot model data were compared to those of healthy children.

CLINICAL SIGNIFICANCE

It is important to take the abnormalities on hemiplegics' unaffected side into account, and to be able to differentiate between primary problems and compensations in gait. The sagittal plane abnormalities found can mainly be explained by compensations for leg length differences. The transverse plane changes may be related to pelvic rotation.

METHODS

The foot kinematics of the unaffected foot of 21 children with hemiplegic CP were collected using the Oxford Foot Model [2] and compared to those of 10 healthy children (20 feet). The CP subjects had a mean age of 11.9 years (range 8-17 years); there were 13 males and 8 females; 11 had their right side and 10 their left side affected. 13 of the subjects had not had surgery, 8 had had foot surgery to their affected side. The average leg length difference was 17mm (range 5 to 30mm), as measured in the clinical examination. The healthy subjects had a mean age of 9.7 years (range 7-16years); there were 2 males and 8 females.

Foot kinematic parameters from the static trial and a single, representative dynamic trial were calculated for each subject. The results of the 2 groups were compared using a student t-test (2-tailed, independent).

RESULTS

In the static standing trials, there is increased knee flexion ($p=0.002$) and increased dorsiflexion of the hindfoot ($p=0.001$) and forefoot ($p<0.001$) in relation to the tibia.

In the dynamic walking trials, there is increased knee flexion at loading response ($P=0.003$). There is decreased maximum plantarflexion of the hindfoot ($p<0.001$) and forefoot in relation to the hindfoot ($p=0.033$) and tibia ($p<0.001$), that is a reduced third rocker. There is increased maximum dorsiflexion of the forefoot in relation to the tibia in swing ($p=0.001$) and stance ($p=0.014$) and of the hindfoot in swing ($p=0.010$). In the transverse plane, there is increased average external tibial rotation ($p<0.001$) and increased internal rotation of the hindfoot ($p=0.033$).

DISCUSSION

Foot kinematics of the unaffected foot of hemiplegic cerebral palsy subjects show some significant differences in the sagittal and transverse planes compared with healthy children. There are no consistent differences in the coronal plane.

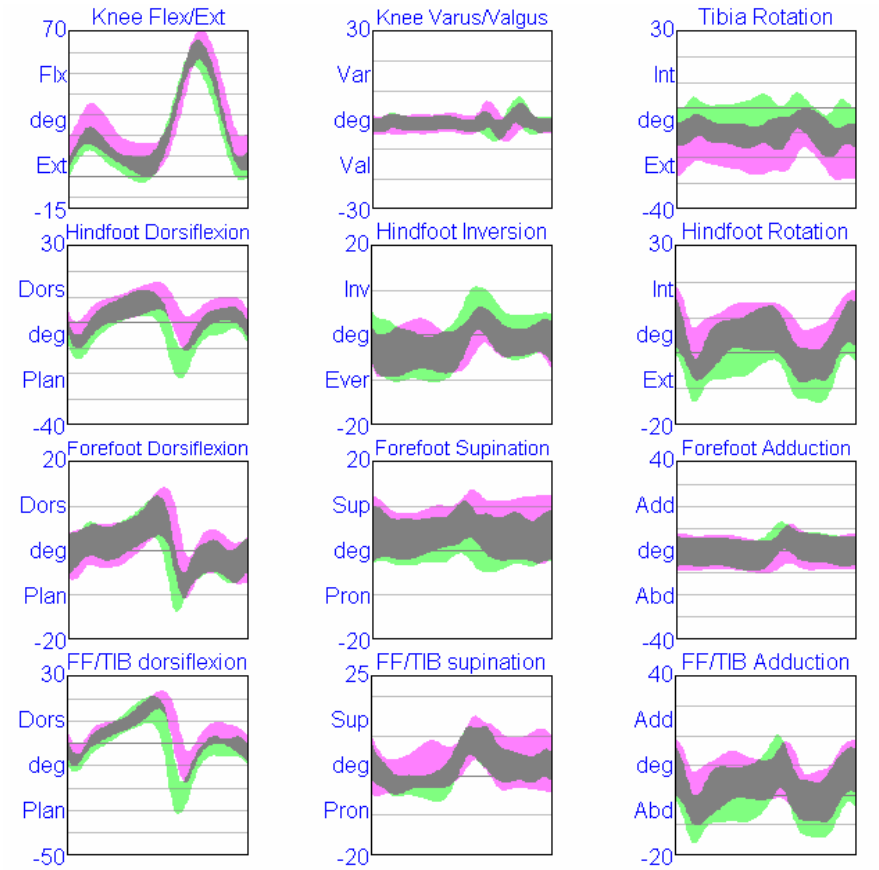

Figure 1: Average foot kinematic data from healthy (green) and hemiplegic unaffected side (pink)

In the standing trials, the increased knee flexion could be explained by the leg length discrepancy. The increased dorsiflexion of the hindfoot and forefoot in relation to the tibia would also correspond with this.

In the walking trials, the increased knee flexion at loading response could also be explained by the leg length discrepancy as this would allow the subject to keep their affected foot flat on the floor at this time. The increased dorsiflexion of the hindfoot and forefoot in relation to the tibia during swing is thought to be a strategy used to aid clearance of this leg. The increased dorsiflexion of the forefoot in relation to the tibia in stance would correlate with the leg length difference and increased knee flexion. The reduced third rocker due to reduced hindfoot plantarflexion could perhaps also relate to leg length as there is no need to plantarflex as much to achieve push-off, or may instead by due to reduced walking speed (the hemiplegic subjects walked at an average of 86% of normal). In the transverse plane, it is possible that the external tibial rotation is a compensation for pelvic protraction and the hindfoot rotation is then a compensation for the tibial rotation.

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Comparative motion of the foot during walking for men, women and children

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Introduction

There are numerous studies in the literature comparing motion of the legs in adults and children during walking. However, very little is known about the differences in these populations for motion within the foot. Motion of the foot in children has previously been reported using the Oxford Foot Model (OFM) [1]. The aim of this study was to compare this to adult males and adult females, and to assess the differences in these populations.

Clinical Significance

It is important to know normal variation within adults and children in order to determine whether a clinical case falls within or outside of this range. In addition, understanding basic differences between different age and gender groups can help in appreciating normal development, and thus distinguishing this from natural history of any particular pathology.

Methods

54 subjects with no known pathology affecting the lower limbs were recruited. Demographics for these subjects are summarised in Table 1.

	Children	Adult Female	Adult Male
n	10	22	22
Age (yrs)	9.7± 2.7	24.7± 3.3	28.2± 7.2
Height (m)	1.4± 0.14	1.7± 0.07	1.8± 0.06
Weight (kg)	31.6± 11.0	61.4± 8.0	71.8± 12.6

Table 1: Demographics for subjects

Each subject had 41 markers attached to both legs and feet by the same assessor, according to the OFM [1]. Subjects were instructed to walk at their normal walking pace along a 10m level walkway. Data were collected with a 12 camera Vicon MX system (Oxford, UK). Clinically relevant variables were selected for analysis and compared between the 3 groups with a one-way analysis of variance (ANOVA).

Results

Results presented in Table 2 summarise significant differences between the groups. Green indicates an increase in the first group compared to the second group, while red represents a decrease. An example of comparison graphs for the OFM (adult males and children) is presented in Figure 1.

Discussion

Results presented here are consistent with previously reported trends. For example, knee valgus alignment was greater in children than adults, and greater in females compared to males. In addition, children's feet were found to be more flexible than adults (in both the sagittal and

	Children /Females	Children /Males	Females /Males
Average Knee Valgus	0.01	<0.01	<0.01
Average Knee Ext Rotation	<0.01	<0.01	
Range HF Dorsiflexion			0.02
Max HF Dorsiflexion Stance	0.02	0.01	
Max HF Dorsiflexion Swing	<0.01	<0.01	
Max HF Plantarflexion			0.04
Range FF Dorsiflexion		<0.01	<0.01
Max FF Dorsiflexion Stance	<0.01	<0.01	
Max FF Dorsiflexion Swing	0.01	0.02	
Average FF Adduction	0.01	<0.01	
Range FF adduction		<0.01	0.02
Average Hallux Valgus			0.03

Table 2: HF = Hindfoot relative to tibia, FF = forefoot relative to hindfoot. P values are listed in the table.

These differences between poplulations should be taken into account when assessing pathology and natural history in any patient.

transverse planes). The combination of reduced hindfoot dorsiflexion, and increased forefoot dorsiflexion in children indicates a flattened arch compared to adults.

Other results indicate that female feet are more flexible than male feet. Women have an increased third rocker at the hindfoot, and more of a varus alignment of the hallux.

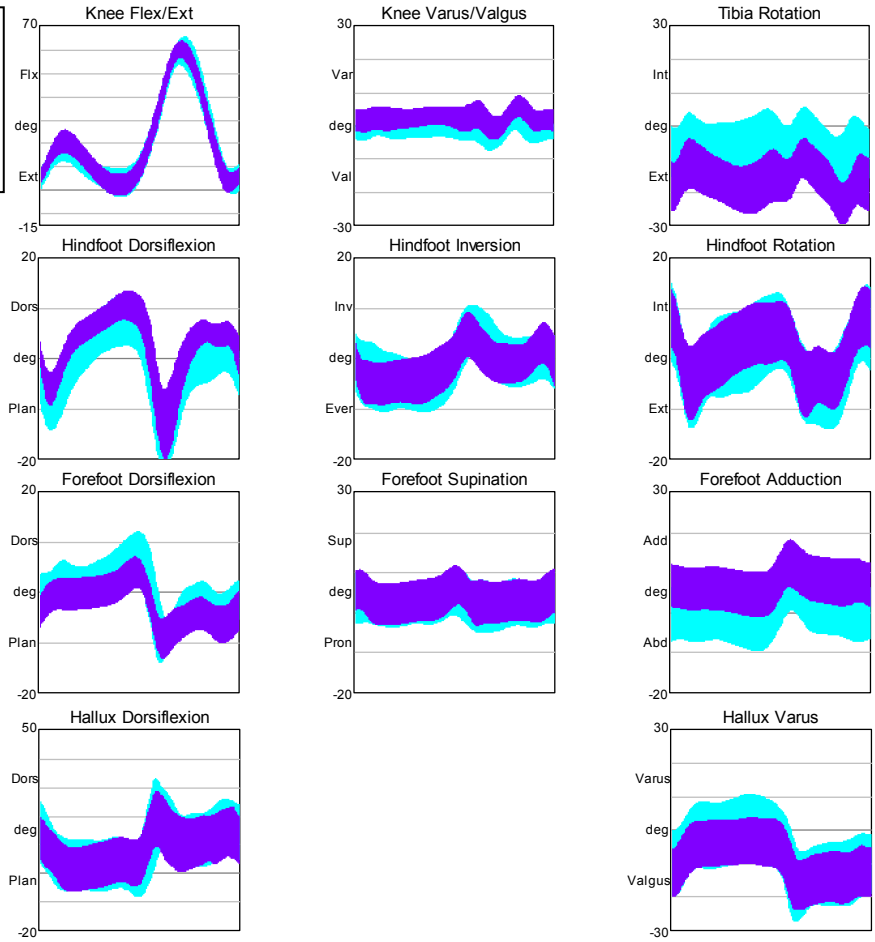

Figure 1: Each graph represents 1 gait cycle (average ± 1SD for each population). Purple = adult males, light blue = children

References

1. Stebbins J, et al. Gait & Posture, 23(4), 401-10, 2006

A COMPARISON OF MULTI-SEGMENT FOOT MOTION DURING OVERGROUND AND TREADMILL WALKING

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INTRODUCTION

Treadmill walking has several advantages during gait evaluation compared to overground walking, including multiple consecutive strides, and ability to control changes in surface incline and speed. Alton et al reported that only sagittal plane hip motion and cadence were different between the two conditions. Riley et al found similar kinematic patterns between overground and treadmill walking in healthy adults, with all differences in peak motion less than 3°. Specifically, they found no differences in ankle plantarflexion/dorsiflexion between conditions. However, a comparison of multi-segment foot kinematics between treadmill and overground walking has not been previously reported.

CLINICAL SIGNIFICANCE

Treadmill walking allows for the design of controlled protocols to evaluation a subject's ability to adapt to different inclines or speeds. Treadmill walking protocols during gait analysis also provides the ability to collect multiple strides in a small capture volume, over a shorter period of time, with less variability in speed between trials.

METHODS AND PROCEDURES

Ten adults, with average age 23.9 ± 5.5 yrs, underwent gait analysis with IRB informed consent. Each subject was instrumented with a modified Helen Hayes marker set, and an eight marker multi-segment foot marker set (Tulchin, 2004). Marker trajectories were collected using a VICON motion capture system at 120Hz and filtered with a Woltring spline with MSE of 10, while subjects performed overground barefoot walking at a self-selected walking speed. Subjects were then asked to complete a treadmill protocol walking at 2.0mph to 4.0mph, in increments of 0.5mph (results not shown.) The speed which most closely matched their self-selected walking speed during the overground condition was used for comparison. Two-tailed, paired t-tests were used to assess differences in multi-segment foot kinematics between conditions.

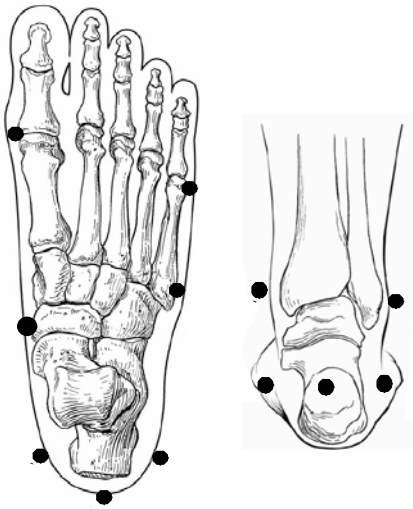

Figure 1: Skin marker placement for the foot kinematic model.

RESULTS

The 2.5mph treadmill speed was selected for six subjects and 3.0mph was used for the remaining subjects. There were statistically significant differences between conditions for maximum Hindfoot plantarflexion ($p=0.04$), maximum Forefoot plantarflexion ($p=0.01$) and maximum Forefoot eversion ($p=0.04$) However, the average change was less than 3º for all three variables, which was not deemed to be clinically significant, Table 1. In addition there was no statistical difference between conditions for ankle sagittal plane motion using the typical lower extremity marker set.

Table 1: Change in kinematic variables were not considered clinically different. mean ± stdev

	Overground	Treadmill
Max HF PF	-16.7° ± 9.2°	-14.2° ± 7.2°
Max FF PF	-15.1° ± 4.7°	-13.3° ± 4.2°
Max FF Ever	-2.4° ± 4.0°	-0.6° ± 3.6°

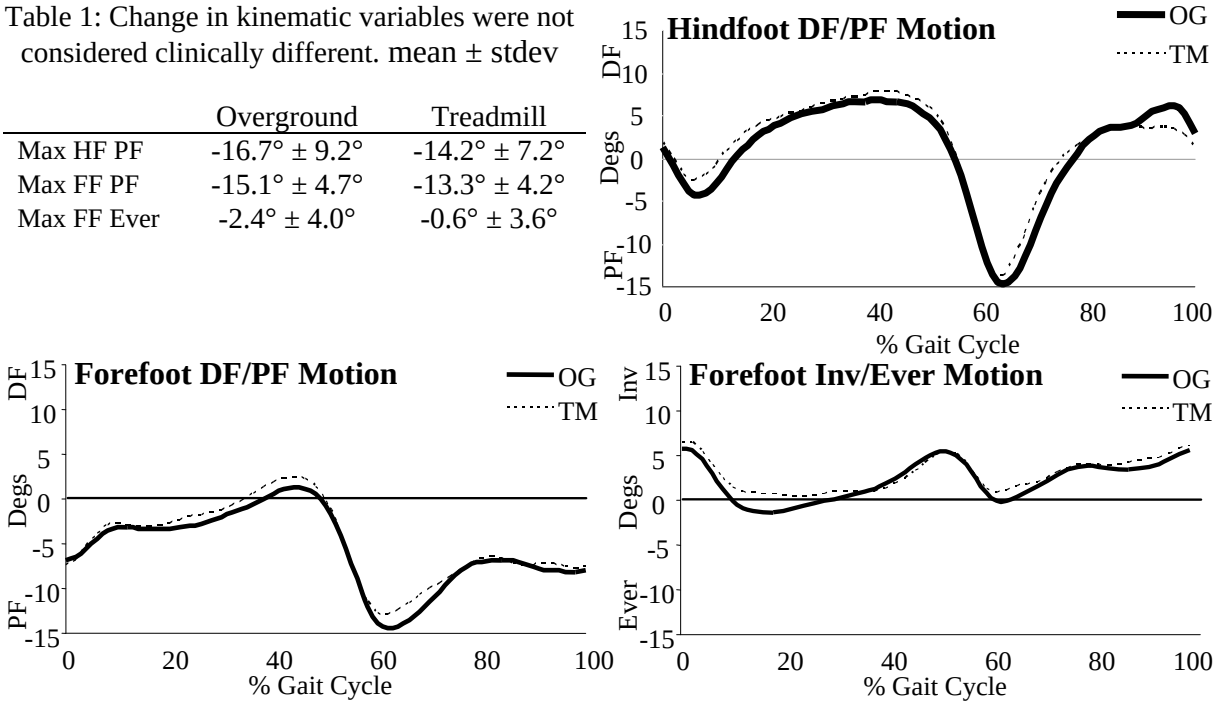

Figure 2: Ensemble average graphs across subjects show that the maximum difference between overground and treadmill walking was less than 3º.

DISCUSSION

No clinically relevant differences were seen in multi-segment foot motion between overground and treadmill walking at comparable speeds. Results from the current study are similar to that of Riley, et al, with all kinematic changes less than 3°. The treadmill can provide a means for easier collection of foot models, particularly in laboratories where the reconstruction of multiple small marker foot data is problematic. It allows for the collection of multiple, consecutive strides within a small capture volume, permitting a seemingly simpler camera placement.

More importantly however, multi-segment foot kinematics can be used in conjunction with treadmill walking to better define foot mechanisms during ambulation at various speeds and inclines. The use of such protocols on subjects with foot pathologies will enable researchers to correlate foot deformity with associated changes in foot motion. In subjects with mild foot pathologies, gait disturbances may not be identifiable during level walking at a self selected speed. Gaining the ability to evaluate subjects at multiple speeds and surface inclines may elicit changes in foot mechanics resulting from limitations in foot motion. In addition, ambulation during daily living in the community often involves modulating speed and negotiating ramps. Treadmill testing can provide additional insight into the mechanisms of foot motion during these tasks.

REFERENCES:

1) Alton, et al, *Clinical Biomechanics*, 13:434-440, 1998.
2) Riley, et al, *Gait and Posture*, 26:17-24, 2007.
3) Tulchin, K and Haideri, N, (2004) *Proceed 8th annual ISB-3D*, p:25-28.

Time	Friday, March 13th, 2009
7:00 –8:00 am	Breakfast –*Foyer* (Registration 7:00 am –9:00 am)
7:00 –8:00 am	Sunrise #3 –*Room 4* "Analyzing Gait Analysis: A New Perspective on Clinical Interpretation" Richard Baker
8:15 –9:15 am	Keynote Address #3 –*Rooms 4–7* "Dystonia, Spasticity, and the Anatomy of Abnormal Reflexes" Terence Sanger, MD, PhD, Stanford University 1 CME/CEU Credit
9:15 –10:15 am	Podium Presentations –*Rooms 1–3* *Session 7a Vascular & Neuromuscular Disorders* *Moderators: Kirsten Tulchin, Dennis Matthews* 1 CME/CEU Credit

- Rating Scales for Dystonia in Children with Cerebral Palsy: Reliability and Evaluation of the Burke Fahn Marshden Scale and the Barry Albright Dystonia Scale
  E. Monbaliu, MSc PT; E. Ortibus, MD; F. Roelens, MD; K. Desloovere, PhD Kinesiologist; H. Isselée, PhD PT; P. Prinzie, PhD; P. De Cock, PhD MD; H. Feys, PhD PT
- Parkinson's Disease Influences Kinematic Regularity of a Simple Leg Swing Task
  V. Ivkovic, J. Bloom, B. McNeil, M. J. Kurz
- Kinematic and Kinetic Gait Parameters and EMG Activity in Children with Haemophilia and Age-Matched Controls
  D. Stephensen, W. I. Drechsler, M. Winter, O. M. Scott
- Lower Extremity Joint Kinetics are Altered in Patients with Unilateral Peripheral Arterial Disease
  P. Koutakis, J. M. Yentes, S. A. Myers, J. P. Kaipust, J. M. Johanning, I. Pipinos, N. Stergiou
- Classification of the Gait Patterns of Boys with Duchenne Muscular Dystrophy (DMD) and the Relationship to Function
  S. S. Thomas, MA; C. E. Buckon, MS; A. Nicorici, BS; A. Bagley, PhD; C. M. McDonald MD; M. D. Sussman, MD
- Breathing Pattern in Muscular Dystrophy
  M. Romei, M. G. D'Angelo, G. P. Comi, S. Bonato, D. Colombo, A. C. Turconi, A. Aliverti, N. Bresolin

9:15 –10:15 am	Podium Presentations –*Rooms 4–7* *Session 7b: Case Studies* *Moderators: Sylvia Ounpuu & Mike Aiona* 1 CME/CEU Credit

- Comparison of Ground Reaction and Articulated Ankle Foot Orthoses in a Child with Lumbosacral Myelomeningocele
  K. Kane, BScPT; J. Barden, PhD
- "If Everything is Fine, Why Do I Keep Tripping?" A Case Study
  S. Sohrweide, PT; T. Novacheck, MD
- Gait Analysis with Dynamic EMG for Decision-Making in an Adult with a Foot Drop of Unknown Etiology: A Case Study
  V. J. Eberly, S. J. Mulroy, V. C. Eberly, J. Perry
- Gait Analysis to Determine Changes in Gait Pattern due to Fatigue in a Child with Cerebral Palsy Spastic Diplegia
  E. M. Godwin, PT, PhD PCS; C. R. Spero, MD
- Case Study: MH Pre and Post Single Event Multilevel Surgery - An Illustration of the Use of the Movement Analysis Profile to Monitor Progress
  P. Thomason, R. Baker, K. Graham
- Gait Analysis for Treatment Decision-Making in Persons with Cerebral Palsy
  S. Õunpuu, MSc; C. Nomides, DPT; M. Westwell, MS, PT; J. Zahradnik, MD

10:15 –10:45 am	Break/Room Change
10:45 am –12:00 pm	Podium Presentations –*Rooms 1–3* *Session 8a: a. Rehabilitation b. EMG* *Moderators: Amy Bodkin, Michael Orendurff* 1 CME/CEU Credit

- Resistance Training Alters Joint Moments and Powers in Multiple Sclerosis Patients Independent of Disease Severity
  J. M. Huisinga, M. L. Filipi, N. Stergiou
- Effects of Locomotor Treadmill Training on Gait Symmetry in Acute Stroke
  S. Coleman, K. McCain, F. Pollo
- The Effect of Body Weight Support Treadmill Training on Paretic Leg Contribution in Hemiparetic Walking in Persons with Chronic Stroke
  E. N. Ozimek, S. Combs, E. L. Dugan
- Surface EMG Does Not Accurately Reflect Rectus Femoris Activity Throughout the Gait Cycle: The Influence of Vasti Crosstalk in Multiple Gait Conditions
  K. M. Barr, A. L. Miller, K. B. Chapin
- Electromyography is Better in Quantifying Muscular Demands Than Joint Moments When Co-Contraction Exists
  C. J. Chen, L.-S. Chou
- Dynamic Stability of EMG Patterns in Young and Elderly Gait
  H. G. Kang, J. B. Dingwell

10:45 am –12:00 pm	Podium Presentations –*Rooms 4–7* *Session 8b: a. Support Surfaces b. Upper Extremity* *Moderators: Nessie Haideri, Roy Davis* 1 CME/CEU Credit

- The Influence of Heel Height on Utilized Coefficient of Friction During Walking
  M. G. Blanchette, MS; C. M. Powers, PhD, PT
- Effect of Surface Incline on Multi-Segment Foot Kinematics
  K. Tulchin, MS; M. Orendurff, MS
- Support Surface and Visual Field Translations Affect Variability and Dynamic Stability of Walking Differently
  P. M. McAndrew, J. B. Dingwell, J. M. Wilken
- Do Humans Optimally Exploit Redundancy to Control Step Variability During Walking?
  J. B. Dingwell, J. John, J. P. Cusumano
- Shoulder Demands Across Manual Wheelchair Activities
  M. M. B. Morrow, W. J. Hurd, K.-N. An, K. R. Kaufman
- Grip Force Production in the Assisting Hand on Performance of a Bimanual Task: Applications to Hemiplegic Cerebral Palsy
  J. Valvano, J. Davis, N. Denniston, T. Boulter, T. Nicklas, and J. J. Carollo
- Upper Extremity Kinetics and Outcomes Assessment During Crutch-Assisted Gait in Children with Myelomeningocele
  B. Slavens, P. Sturm, G. Harris

12:00 –12:15 pm	Awards Presentation –*Rooms 5 –7*

12:30 –3:00 pm	Technical Symposium–*Rooms 1-3* "Biomechanical Models: As Simple as Possible, but no Simpler" 12:30 –1:45 Title: Introduction to DW and MMS Presenters: *Scott Delp PhD and Art Kuo PhD* 1:45 – 2:30 Title: Research presentations: Applications to clinical practice Presenters: *Rick Neptune PhD and Max Donelan PhD* 2:30 –3:00 Panel Discussion and responses to participant's questions Panel discussants: *Scott Delp PhD, Max Donelon PhD, Roger Kram PhD, Art Kuo PhD, Rick Neptune PhD, Caroline Stewart PhD, Marjolein van der Krogt MsC.* 3 CME/CEU Credits
3:15 –5:45 pm	Clinical Symposium –*Rooms 1–3* "Evaluation of Dystonia on the Motion Analysis Laboratory: Application to Cerebral Palsy" 3:15 – 3:45 Title: Dystonia in children with cerebral palsy: Diagnosis and intervention. Presenter: *Terence Sanger MD, PhD* 3:45 – 4:25 Title: Clinical assessment of dystonia in the motion analysis laboratory: Visual assessment tools and observational assessment during motion analysis. Presenters: *Kim Wesdock PT and Cammie Beattie PT* 4:25 – 5:00 Title: Motion analysis tools to differentiate dystonia only, spasticity only and dystonia mixed and spasticity. Presenters: *Diane Damiano PhD and Katharine Alter MD* 5:00 – 5:45 Title: Case Examples and Panel Discussion Panel discussants: *Katharine Alter MD, Diane Damiano PhD, Susan Rethlefsen PT, Dennis Matthews MD, Freeman Miller MD and Terence Sanger MD* 3 CME/CEU Credits
6:30 –7:30 pm	Happy Hour / Game Hour – Rooms *1–3/Foyer*
7:30 –11:30 pm	Gala Dinner – Sponsored by AMTI Rooms *4–7* Entertainment by Wild 6

Rating Scales for dystonia in children with cerebral palsy: reliability and evaluation of the Burke Fahn Marshden Scale and the Barry Albright Dystonia Scale

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Introduction: The past two decades, the interest in pediatric movement disorders, such as spastic cerebral palsy (CP) has increased remarkably.[1] However dyskinesia (dystonia and choreo-athetosis) is underreported in contrast with the spastic CP group.[2] The cause can be found in the complexity of the movement disorder and the former confusing definition and classification of dyskinesia in CP. No standardized scales are known to classify choreo-athetosis or to grade its severity.[1] Dystonia has been graded most often with the Burke-Fahn-Marshden dystonia rating scale (BFMS)[3] developed for adults with primary dystonia. The Barry Albright Dystonia Scale (BADS) [4] was modified from the BFMS for use in children with CP. Until now, the reliability of both scales in dystonic CP is not yet fully investigated. In addition a critical evaluation of both scales according to the new proposed definition and classification of CP[5] and the International Classification of Functioning, Health and Disability model (ICF)[6] is needed. The aims of this study were (1) to evaluate the reliability of the BADS and BFMS and (2) to evaluate the content of both scales in children with dystonic CP.

Clinical significance: A reliable and valid rating scale for the evaluation of dystonia in CP is of major importance to increase pathology insights according to the most recent definition and classification of CP and to evaluate therapeutic interventions for this population.

Methods: Ten patients with dystonic CP participated in this study (mean age = 14.0 years, SD = 6.14, range 4.5 to 25 years). Gross Motor Function Classification levels ranged from 1 to 5. To evaluate the inter rater reliability, three raters independently scored ten videotaped patients with both the BADS and BFMS. For test-retest reliability, a second videotape for five patients was made one week after the first videotape and was scored by the three raters with a time interval of 4 weeks between the scoring of test and retest. Reliability was assessed using Intraclass Correlation Coefficients (ICC). Internal consistency was assessed for one rater with Cronbach's alpha. The Standard Error of Measurement (SEM) and the Smallest Detectable Difference (SDD) were calculated for both scales.

Results: The BFMS and BADS revealed a high level of interrater reliability for the total scores, with an ICC of respectively 0.89 and 0.87. Interrater reliability for the nine individual BFMS item scores ranged from 0.32 to 0.93 and for the eight individual BADS item scores ranged from 0.40 to 0.76. Test-retest reliability of the total scores for the three raters ranged from 0.86 to 0.93 (BFMS) and 0.78 to 0.98 (BADS). Both scales, BFMS and BADS, were found to have a high level of internal consistency with a Cronbach's alpha of respectively 0.93 and 0.90. The SEM and SDD was 6.20% and 14.52% of the total score range for the BFMS and 2.11% and 6.58% for the BADS. Inspection of the individual data shows a limited spread in the obtained scores and suggests a lack of sensitivity of the scales. Analysis of both scales revealed that the BFMS and BADS definitions of dystonia have insufficient accordance with the recent cerebral palsy definition. Also, the scales intermingle the body function and activity level according to the ICF model. In both scales, there is no distinction in dystonia between rest and activity nor between the amplitude and duration. There is also no possibility to differentiate dystonia between proximal and distal limbs.

Discussion: This study indicates that the BADS and the BFMS are reliable scales with a good to excellent interrater and test-retest reliability for the total scores. Reliability for the individual items scores was lower but acceptable for most items of both scales. SEM and SDD were within the expected values for clinical rating scales. However, critical analysis of both scales revealed several limitations and hereby confirm the need for a measurement tool for a valid, reliable and sensitive measurement scale to document the degree of dystonia in CP, as suggested by Butler[7]. In accordance with the classification that subdivides dyskinesia in dystonia and choreo-athetosis, and in view of the fact that clinical presentation of both dystonia and choreo-athetosis at the same time is common, it is recommended to evaluate dystonia and choreo-athetosis within one scale taking into account the new definition of CP.

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Parkinson's Disease Influences Kinematic Regularity of a Simple Leg Swing Task

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Introduction

The regularity of the variations found in human biorhythms have been linked to age and disease states [1-2]. For example, the structural variations that are found in a rhythmical finger force task are less regular in the aged and individuals with Parkinson's disease (PD) [1]. These observed changes represent an inability of the nervous system to integrate the motor-sensory information for the maintenance of a regular motor pattern [1]. Currently, it is unknown if the changes in regularity of the structural variations persist while performing a rhythmical motor task with the lower extremity, and if the frequency of the rhythmical pattern influences the regularity. The aim of this study was to identify neuromechanical factors associated with changes in kinematic regularity of the structural variations present in a simple rhythmic leg swing task.

Clinical Significance

Currently there is a need for effective biomarkers for the diagnosis and management of PD [3]. Our results indicate that PD and not aging results in changes in the regularity of the structural variations present in a simple leg swing task. We suggest that the outcomes presented here can be used as a spring-board for the developemnt of a biomarker that can differentiate between the symptoms associated with PD, and changes in the motor system due to the normal aging process.

Methods

A leg swing task was used to explore the neuromechanical factors that govern the regularity of the structural variations present in the motor output (Figure 1). Young (n=9, Age=19.9±1.3 yrs.), aged (n=9; Age= 74.7±6.1 yrs.), and individuals with PD (n=9, Age=73.4±6.6) participated in the study. PD individuals had a Hoehn-Yahr score between 2 and 3, and performed the experiment while "on" levodopa therapy. Participants swung their leg at their preferred pendular frequency, and frequencies that were 20% faster and slower. Equation 1 was used to calculate the period of the prescribed leg-swinging task.

$$T = 2\pi\sqrt{\frac{I}{mg\ell}} \qquad \text{Equation 1.}$$

where T is the period, I is the inertia of the leg, m is mass, g is gravity and ℓ is the length of the leg. Based on the calculated period, a metronome was used to help the participants maintain the prescribed

Figure 1. Participant performing the leg swing task. A goniometer was used to monitor the limb's angular position.

frequency (1/T). Participants swung their leg for two minutes at each frequency. Approximate Entropy (Equation 2) was used to measure the regularity of the structural variations present in the leg swing kinematic data collected by an electronic goniometer (100 Hz) [1].

$$ApEn(m,r) = \ln\left(\frac{C_m(r)}{C_{m+1}(r)}\right) \quad \text{Equation 2.}$$

where $C_m(r)$ is the number of data points of length m that are similar, $C_{m+1}(r)$ is the number of data points of length m+1 that are similar, and r is the similarity criterion. For this investigation, m was 2 and r was 20% of the standard deviation of the time series. A lower ApEn value indicates a greater regularity in the structural variations present in the leg kinematics.

Results

The main effects for frequency were significantly different ($p<0.001$; Figure 2), and less regular structural variations were associated with faster frequencies. The group main effects were significantly different ($p<0.05$; Figure 2). The regularity of the structural variations present in the leg swing kinematics of the aged group were not significantly different than the young at all frequencies ($p>0.05$). The regularity of the structural variations present in the PD group's leg swing kinematics were significantly different than the aged and young at all frequencies ($p<0.05$; Figure 2).

Discussion

Our results indicate that the regularity of the structural variations present in a leg swing task is influenced by the frequency of the movement pattern. However, changes in the regularity of the structural variations present at each frequency are not influenced by age but were influenced by PD. This infers that the basal ganglia plays a large role in controlling the regularity the lower limb rhythmical pattern. Changes in the regularity of the leg-swinging pattern may offer a new biomarker for the assessment and management of PD. Future investigations should explore if the regularity of the limb-swinging pattern is influenced by the severity of the disease.

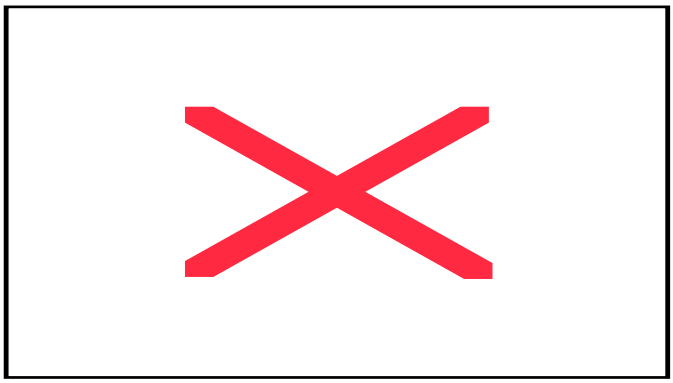

Figure 2. Kinematic regularity measured by Approximate Entropy (ApEn) at three frequencies (low, pendular, high) in three groups of participants (young, aged, Parkinsonian).

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Kinematic and kinetic gait parameters and EMG activity in children with haemophilia and age-matched controls

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Introduction & Clinical Significance: We have recently reported significant changes in the biomechanical parameters of gait in children with haemophilia which imply that lower limb function is more impaired than parallel clinical evaluations indicate [1]. The aim of this study was to provide an extended analysis of sagittal plane lower limb biomechanics together with changes in EMG activity during walking, between a control group of normal age matched children (Gp1) and children with haemophilia and a history of haemorrhage into the ankle joint (Gp2). **Methods:** A ten camera Vicon 612 Motion Analysis System, (Oxford Metrics Ltd, UK), 2 Bertec Force Platforms (Model 4020 H, MIE Ltd, Leeds, UK) and a Zerowire multichannel wireless EMG system (Aurion, Milan Italy) were used to collect kinematic, kinetic and surface EMG data from 21 boys in Gp1 (10.14 ± 1.76 yrs) and 21 boys in Gp2 (10.80 ± 1.82 yrs). Integrated EMG signals for vastus lateralis (VL), biceps femoris (BF), tibialis anterior (TA) and lateral gastrocnemius (LG) during the gait cycle were normalised using isometric MVC. Between group mean differences for all parameters during initial double support, single support, terminal double support and swing were analysed using one way ANOVA. **Results** There was an increase in angle of knee flexion ($P<0.05$) throughout the gait cycle in Gp2 together with reduced (Gp1, -0.42Nm/kg ± 0.18; Gp2 -0.31 Nm/kg ± 0.17; $P<0.05$) knee extensor moment at initial double support and reduced hip extensor moment (Gp1, -0.11 Nm/kg ± 0.23; Gp2, 0.05 Nm/kg ± 0.20; $P<0.05$) during terminal double support.

The shape of the vertical ground reaction force (GRF) was similar in both groups but the second vertical GRF peak during terminal double support was significantly reduced ($P<0.05$) in Gp2 prior to toe-off (Fig 1). Significantly increased ($P<0.05$) levels of EMG activity of LG during both single and terminal double support were found in Gp2 suggesting that a critical phase in the haemophiliac children occurs at toe off and terminal stance (Fig 1).

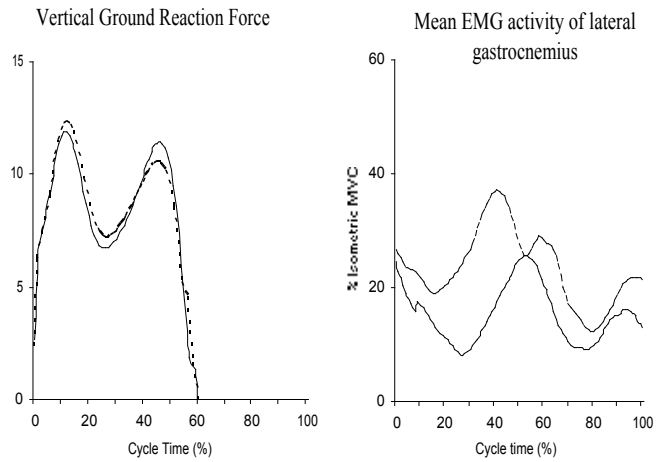

Figure 1. Vertical ground reaction force and mean integrated EMG recordings (normalised using isometric MVC) from lateral gastrocnemius during gait. Continuous and dashed lines represent Grp 1(n=21) and Grp 2 (n=21), respectively.

Discussion: The results suggest changes in ground reaction forces and EMG activity patterns of lateral gastrocnemius are present in young haemophiliac children with a target ankle joint in addition to previously observed kinematic and kinetic changes during level walking.

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Acknowledgments: The Haemophilia Society UK for a supporting grant and the children and families who took part in this study

LOWER EXTREMITY JOINT KINETICS ARE ALTERED IN PATIENTS WITH UNILATERAL PERIPHERAL ARTERIAL DISEASE

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INTRODUCTION

Peripheral arterial disease (PAD) affects over 8 million people in the US and is a manifestation of atherosclerosis leading to decreased blood flow to the legs. The result is ischemic pain (claudication) that is induced by physical activity and results in diminished ability to walk [3].

CLINICAL SIGNIFICANCE

Minimal research work has been done to determine the gait handicap of PAD patients. Recently we have found that patients with bilateral PAD have altered ground reaction forces and joint kinetics and kinematics as compared to controls [1, 2]. The present investigation extends this work to the evaluation of joint kinetics in patients with unilateral PAD. We hypothesized that the joint kinetics of unilateral PAD patients would be different between the affected and the non-affected limbs and as compared to healthy matched controls. We also hypothesized that these differences would become worse after the onset of claudication pain.

METHODS

Thirteen, unilateral PAD, older adults (age: 61.69±10.53yrs, mass: 84.65±20.24kg, height: 1.72±0.08m), and eleven matched healthy controls (age: 66.27±9.22yrs, mass: 77.89±10.65kg, height: 1.74±0.08m) walked over a force platform (Kistler Instruments, Switzerland) to acquire kinetics, while joint kinematics were recorded simultaneously with a 8-camera motion capture system (Motion Analysis, Santa Rosa, CA). Five trials were collected for each limb in the pain free (prior to the onset of claudication) and pain induced conditions. The pain free condition was acquired first and one minute rest periods were required between each walking trial. After completion of the pain free condition, pain was induced by having subjects walk on a treadmill at 10% grade and 0.67 m/s. At the onset of pain, the pain induced condition walking trials were collected with no rest between trials. The data were combined to calculate joint moments and powers during the stance phase of gait. A 2X2 fully repeated measures ANOVA (Limb by Condition), was used to identify differences between the group means of the peak joint moments and powers of the PAD patients. Independent *t*-tests were also used to compare the group means from both conditions and both limbs of the PAD patients with the healthy controls.

RESULTS

Results are presented in Table 1 below. Only significant results are reported.

DISCUSSION

Our results point to several important conclusions. First, in unilateral PAD patients even the non-affected limb is actually affected. This was evident by the lack of significant main effects for the limb factor in the PAD patients and the similarity of statistical findings with respect to powers as compared with controls. Second, claudication significantly affected several parameters; specifically, decreased ankle plantarflexor moment and power and increased knee power at early mid-stance. The additional decrease at the ankle plantarflexor seems to be a commonality between unilateral and bilateral PAD patients as we have observed in our previous work [1]. Third, this study revealed several functional deficits in unilateral PAD patients such as an inability of the knee flexors to eccentrically control weight acceptance after heel strike, weakness of the hip and knee extensors to concentrically support the trunk in early stance, and weakness of the ankle plantarflexors in late stance to concentrically progress the body forward.

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Table 1: Group means (±SD) from peak joint moments and powers.

	Pain free Non-Affected Limb	Pain free Affected Limb	Pain induced Non-Affected Limb	Pain induced Affected Limb	Control
Ankle Dorsiflexor Moment (N*m/kg)	-0.38±0.20	-0.29±0.13	-0.42±0.30	-0.23±0.15*	-0.36±0.09
Ankle Plantarflexor Moment (N*m/kg)	1.32±0.16	1.18±0.25*	1.27±0.18†	1.11±0.27*†	1.31±0.28
Knee Extensor Moment (N*m/kg)	0.61±0.30	0.58±0.27*	0.69±0.41	0.59±0.37*	0.82±0.18
Ankle Plantarflexor Power Late Stance (w/kg)	2.65±0.9*	2.49±0.46*	2.39±0.67*†	2.05±0.59*†	4.00±0.88
Knee Power Early Stance (w/kg)	-0.52±0.4*	-0.36±0.2*	-0.75±0.74	-0.36±0.32*	-0.73±0.22
Knee Power Early Mid-stance (w/kg)	0.31±0.2*	0.25±0.25*	0.41±0.28*†	0.26±0.31*†	0.62±0.25

Note: * $p < 0.05$, significant differences with healthy controls
† $p < 0.05$, significant main effects for pain
No significant main effects for limb
No significant interactions

CLASSIFICATION OF THE GAIT PATTERNS OF BOYS WITH DUCHENNE MUSCULAR DYSTROPHY (DMD) AND THE RELATIONSHIP TO FUNCTION

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Introduction

Duchenne muscular dystrophy (DMD) is an X-linked recessive disease of muscle characterized by a progressive loss of functional muscle mass. Gait deviations are coupled with the development of contractures at the hip, knee and ankle and boys with DMD lose the ability to walk between the ages of 8-12 years. Corticosteroids have been shown to reduce the expected loss of muscle strength and extend the time that ambulation is maintained; however, minimal information exists regarding the relationship between gait deviations and function in boys with DMD. The gait deviation index (GDI) was developed to quantify the degree of gait pathology relative to normal and provide a single measure of gait quality[1]. It is composed of fifteen kinematic control features that stratify gait deviations based on severity. The purpose of this study was to characterize gait in boys with DMD using two methods, 1) comparing average gait patterns based on steroid vs steroid naive[2] and 2) utilization of the 15 kinematic control features of the GDI[1] to cluster patterns of deviation, and to determine the relationship between these methods and functional measures commonly used to define disease progression in boys with DMD.

Statement of Clinical Significance

Classification of boys with DMD based on gait deviations, rather than steroid use, may provide insight into patterns of disease progression and the relationship to function.

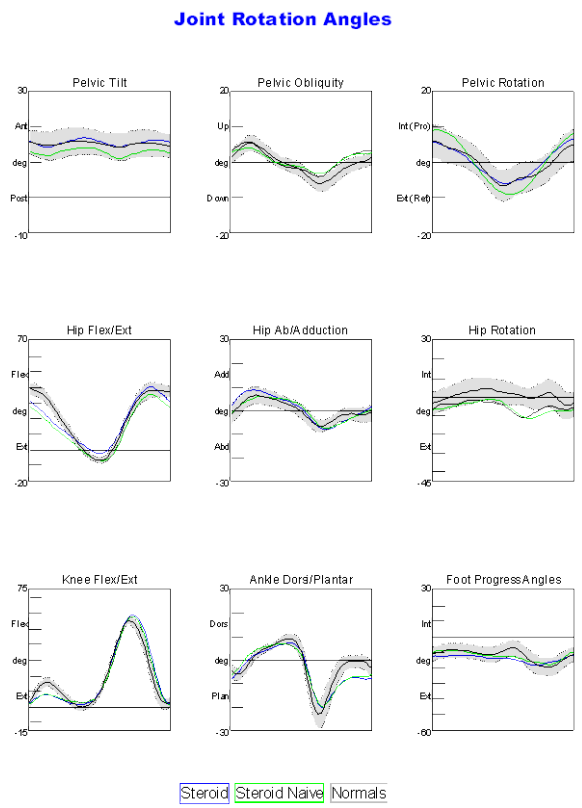
Methods

Forty-one boys with DMD, mean age 8 yrs, 1 mth (range 4-15 yrs) completed a quantitative gait analysis (VICON, Motion Analysis) and assessment of energy efficiency (Cosmed K4b[2] unit), gross motor function (GMFM-66), and timed motor performance (run 10 m, climb 4 stairs). Gait patterns were evaluated using two methods: steroid vs. steroid naïve and by cluster analysis based on a k-means algorithm of the 15 kinematic control features of the GDI. Cluster analysis identified three patterns of gait deviation (figure 2). Independent samples t-test was used to determine whether significant differences in functional variables exist between boys on steroids and boys who are steroid naïve, while a one-way ANOVA was used to determine whether there were significant difference in functional variables exist between the boys in each of the 3 clusters. Significance was set at $p<.05$.

Results

	Steroid Usage		Gait Deviation Clusters		
Variable	**S Naive**	**Steroid**	**Cluster 1**	**Cluster 2**	**Cluster 3**
Number of Boys (N)	17	24	16	15	10
Steroid Naïve (SN) / Steroid (S)			3/13	9/6	5/5
Age (months)	89	103	97	87	111
GMFM-66	71	69	68	73	68
Time to Run 10 meter (s)	6.37	6.06	6.7	5.7	6.1
Time to Climb 4 stairs (s)	4.42	5.85	6.5	4.6	4.0
Velocity (m/s)	.747	.750	.763	.775	.687
NNcost	.357	.342	.356	.315	.387
GDI	75.94	81.57	81.89	83.42	68.72

No significant differences were found between the functional variables for boys on steroids and those who are steroid naïve. When comparing the clusters, only GDI was found to be significantly different between boys in cluster 3 and those in cluster 1 ($p=.049$) and 2 ($p=.027$).



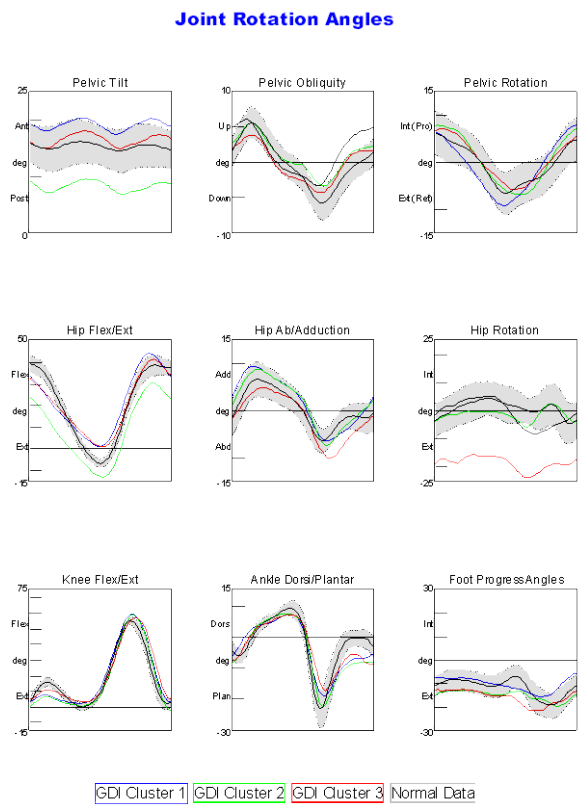


Figure 1: Group means Steroid vs Steroid Naïve Figure 2: Cluster by gait deviation

Discussion

When comparing the gait patterns of the boys with DMD by steroid usage, subtle differences in pelvic kinematics only were noted, with greater normality seen in the steroid group as reflected by the GDI score. Functionally the two groups performed similarly, despite the older age of the steroid group. Cluster analysis did reveal different patterns of gait deviations. In relationship to function, boys with the lowest GDI score (most deviation) have the slowest velocity and the highest energy cost. Boys with the highest GDI (least deviation) have the highest GMFM-66, fastest velocity, and lowest energy cost. The results of this study demonstrate that in boys with DMD, gait pattern deviations are related to function, which may provide further insight into the understanding of the disease progression. Further study is needed to determine whether the gait deviation clusters identified in this study are indicative of a progressive pattern of muscle weakness or reveal differences in the manifestation of DMD. Additionally, a larger sample size is needed to determine the impact of steroid use on the progression of gait deviation patterns and function.

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BREATHING PATTERN IN MUSCULAR DYSTROPHY

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Introduction

Muscular Dystrophy (MD) is a group of degenerative disorders of the skeletal muscle that develop secondary to abnormalities of dystrophin or dystrophin-associated proteins. Most MD follow a progressive clinical course, culminating in respiratory insufficiency. Many previous studies focused on respiratory function of MD patients, but the size of analyzed groups was small, subjects with different MD were considered as a single group and the investigation was conducted mainly through the spirometry [1-5], beside the difficulties due to facial muscular weakness in the use of the mouth piece in some MD patients.

In the present study a large cohort of MD patients was recruited and breathing pattern was quantitatively investigated and described by using Optoelectronic Plethysmography (OEP, [6]). OEP is a technique that starting from 3D coordinates of reflective markers positioned on trunk and acquired by an optoelectronic system allows to calculate ventilatory parameters and chest wall (CW) volume.

Materials and Methods

62 adult patients with MD (27 Limb-Girdle Muscular Distrophy – LGMD, 14 Becker Muscular Distrophy – BMD, 21 Facio-Scapulo-Humeral Distrophy – FSHD) and 20 healthy age and sex matched volunteers (Control Group – CG) were involved in this study (Table 1).

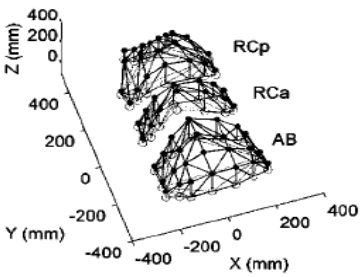

Fig. 1. CW compartments

Table 1. Pathological and healthy groups' data

	N	Sex	Age (years)	Weight (kg)	Height (cm)
LGMD	27	11 F, 16 M	35.2±14.7	63.3±15.8	169.4±8.2
BMD	14	14 M	35.8±12.1	64.7±9.1	169.1±6.2
FSHD	21	9 F, 12 M	40.5±18.3	68.1±12.5	169.1±10.6
CG	20	6 F, 14 M	32±9.2	70.9±14.8	174.9±7.6

Subjects' breathing pattern in supine position was investigated by using OEP (OEPSYSTEM, Bts Bioengineering, Italy) and respiratory function was analyzed by spirometry. 52 reflective markers were positioned on subject's trunk from clavicles to pubis for the calculation of CW volume and its thoraco-abdominal compartments (RC_p: Pulmonary Rib Cage, RC_a: Abdominal Rib Cage, AB: Abdomen, Fig. 1). Subjects were asked first to quietly and spontaneously breath for 3 minute (QB), and then to perform two Slowed Vital Capacity (SVC) with a rest period of 30 seconds.

For QB analysis, starting from CW volume traces, several consecutive breaths (from 10 to 20) were chosen to obtain a single representative normalized breath. For each subject the complete ventilatory pattern was defined and Minute Ventilation (V'_E), Breathing Frequency (RR), Tidal Volume (V_T) and CW compartments percentages were calculated. For SVC analysis, for each subject from the CW volume traces the two SVC were selected and the mean values of inspired and expired Volume (Vin, Vout) and the CW compartments percentages during inspiration and expiration phases $RC_{p,in}$, $RC_{a,in}$, AB_{in}, $RC_{p,out}$, $RC_{a,in}$, AB_{out}) for the two SVC were calculated. For each group and for each parameter of QB and

SVC mean and standard deviation were calculated. Statistical analysis was used to test differences with CG data and between groups.

Results

Spirometry data showed values lower than predicted for all the three groups (Table 2), except for FEV1/FVC that was normal and RV that was higher than normal.

Table 2. . Spirometry data (mean % of predicted values and standard deviation in brackets)

	FVC	FEV1	FEV1/FVC	PEF	TLC	RV	VC
LGMD	76.6%(21.8)	78.3%(21.2)	86.4%(6.7)	72%(17.5)	84.7%(19)	109.7%(34.6)	76.5%(21.9)
BMD	91.1%(17.2)	93.2%(18.6)	85.6%(9.8)	75.8%(20.4)	94.2%(10.1)	109.5%(21.3)	89.2%(18.7)
FSHD	81.1%(17.8)	79.3%(22.2)	81.4%(13)	62.9%(19.5)	88.45%(14.7)	113.3%(29.0)	81.6%(17.8)

During QB no significant differences were found in ventilatory and CW compartments parameters between pathological groups and CG. In SVC trial, dystrophic subjects presented ventilatory parameters and breathing pattern that markedly differentiated from those calculated for healthy subjects (Table 3). In pathological ventilatory parameters, the most evident difference with CG was obtained for Vout_N values. The more compromised were the subjects affected by FSHD: the pattern of their CW compartments noticeably deviated from normality both during inspiration and expiration phase.

Table 3. Ventilatory parameters during SVC (mean values and standard deviations in brackets)

SVC	Vin	Vout	$RC_{p,in}$	$RC_{a,in}$	AB_{in}	$RC_{p,out}$	$RC_{a,in}$	AB_{out}
LGMD	2.7(1.1)	3.5(1.3)	38.6(13)	19.2(8.2)0	42.8(17.2)	39.1(12.2)	27.1(9.1)	34.1(13.9)
BMD	3.3(0.8))	4.2(0.9)	37.5(6.2)	23.8(7.2)	39.8(9.8)	38.8(7.9)	28.4(5.9)	32.5(9.5)
FSHD	2.9(0.9)	3.5(1.1)	33.5(9.8)	16.3(6.5)	51.5(13.1)	37.4(9.4)	20.6(9.3)	42.3(15.3)
CG	3.7(0.6)	4.9(0.8)	42.4(7)	20.3(4.2)	39.6(9.5)	39.6(7.5)	24.2(4.3)	36.5(7.9)

Underlined data were significantly different ($p<0.05$) when compared with CG.

Discussion

Spirometry data revealed that adult MD patients present a mild restrictive respiratory pattern (normal FEV1/FVC but decreased VC, FVC and TLC and increased RV) probably due to respiratory muscle weakness.

The analysis of QB through the OEP, showed normal respiratory function and breathing pattern comparable with healthy subjects. A more demanding task such as SVC revealed in all the three pathological groups a reduced ability to perform a deep expiration, particularly evident for non ambulant patients, probably due to diaphragm weakness. In addition, OEP data demonstrated that FSHD patients had the most abnormal breathing pattern in SVC, especially during the inspiration phase, where the diaphragm and the muscles of abdomen were recruited more than CG and the other pathologic groups. This was probably due to weakness of upper trunk muscles, characteristic of FSHD.

OEP technique demonstrated to be an useful tool even for subjects that are not able to perform spirometry due to facial weakness. Based on OEP data, it was possible not only to evidence and quantify differences in ventilatory parameters between MD groups and healthy subjects, but also to identify a breathing pattern specific for different dystrophies and then to unravel early signs of respiratory deficiency.

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Comparison of ground reaction and articulated ankle foot orthoses in a child with lumbosacral myelomeningocele

Introduction: Physicians, orthotists, and physiotherapists are often required to make challenging clinical predictions about the type of ankle foot orthosis (AFO) that should be fabricated in order to optimize a child's gait mechanics and function. In more complex situations, the efficacy of clinical decision-making may be enhanced by the use of three-dimensional motion capture technology to more objectively quantify important gait variables.

Ground reaction AFOs (GRAFOs), which function to improve the internal knee extension moment during weight-bearing, have been recommended for children with myelomeningocele and a crouched gait pattern. Appropriate alignment of the foot and knee has been cited as a prerequisite for their use, although more specific guidelines are not currently available. This case study illustrates the clinical application of three-dimensional motion analysis technology to quantify kinematic gait variables during the process of orthotic evaluation in an adolescent with myelomeningocele and tibial torsion. It was hypothesized that the use of GRAFOs would reduce the participant's stance phase hip and knee flexion, characteristic of the crouched gait pattern commonly seen in children with this condition.

Clinical Presentation: The participant was a 14-year-old girl with lumbosacral myelomeningocele. She was a community ambulator with the aide of Lofstrand forearm crutches and articulated AFOs. The AFOs were designed to assist ankle dorsiflexion to neutral, as well as to prevent dorsiflexion past neutral. Clinically she demonstrated a crouched gait pattern, along with bilateral external tibial torsion (20º on the left leg and 45º on the right), 10º hip flexion contractures, and increased lumbar lordosis.

Gait Data: A three-dimensional optical tracking (VICON) system and force plate (AMTI) were used to compare static postural sway and lower extremity joint kinematics associated with three gait conditions: barefoot walking, articulated AFOs (with dorsi-assist and dorsiflexion stop), and GRAFOs. Lower extremity joint kinematics for an entire gait cycle were assessed using qualitative comparison of values between the three gait conditions and normal kinematic curves. Mean foot centre of pressure excursion during quiet standing and sagittal plane kinematics during the stance phase of gait were also compared.

The greatest improvements in gait kinematics were observed in the sagittal plane, on the left leg, while wearing the GRAFOs. During these trials, peak stance phase hip and knee extension on the left side were closest to normal values when compared to the other conditions. Peak left knee extension during stance approached normal values in the GRAFO trials (7º flexion) compared to barefoot and articulated AFO trials, in which the knee did not extend past 19.5º of flexion (see Figure 1). The left hip reached 3º of extension during stance in the GRAFO trials, while it remained flexed throughout the articulated AFO (maximum of 8º flexion) and barefoot (maximum of 3º flexion) trials. The GRAFO limited left ankle stance phase dorsiflexion (maximum 5º of dorsiflexion during stance) more effectively than the articulated AFO (maximum 28º). No clear difference between conditions was apparent on the right leg.

The larger degree of tibial torsion present on the participant's right side likely limited the effectiveness of the right GRAFO, resulting in better results for the left leg.

Treatment Decisions and Indications: The data supports the clinical application of computerized gait analysis as an adjunct to observational assessment alone, offering the clinician an opportunity to select the orthosis that will provide the greatest benefit to the client. In addition, these results provide further information about the effects of GRAFOs on gait mechanics in individuals with tibial torsion, supporting and clarifying previous reports (Gage, 2004) that rotational deformities should be considered when prescribing a GRAFO. Based on our results, we suggest that an individual with a 20º external tibial torsion deformity may be considered an appropriate candidate for GRAFO use; however, with more severe deformities, such as those approximating or exceeding 45º, the GRAFO's biomechanical effectiveness may be impaired. Based on our results, GRAFOs were recommended for this girl, and further modifications were carried out on the right GRAFO in an attempt to improve her gait mechanics.

Summary: These results demonstrate that GRAFOs have the potential to improve sagittal plane kinematics in individuals with neuromuscular impairments such as lumbosacral myelomeningocele, thereby reducing the demand for hip and knee extensor activity, and improving gait efficiency. This information is also applicable to children with cerebral palsy, who may similarly demonstrate a crouched posture and tibial torsion. To date, the use of this technology in clinical practice has generally been limited to larger centres; however this case study represents an innovative collaboration between an academically-based laboratory and a rehabilitation centre. As the appreciation for the clinical value of this technology improves, clinicians in our health region are beginning to refer clients for computerized gait analysis.

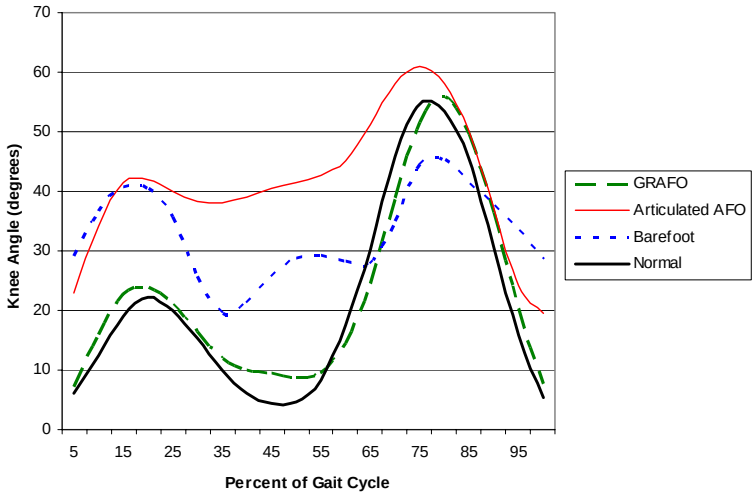

Figure 1. Graph showing sagittal plane kinematics of the left knee for one gait cycle (from initial contact to initial contact).

"If everything is fine, why do I keep tripping?" A Case Study

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Patient History:

The patient is a 69+2 year old female referred to the Center for Gait and Motion Analysis for an initial assessment of her walking, secondary to persisting left foot pain and "toe catching". She has a long history, dating back to 1995, of symptoms in her left lower extremity. She described these symptoms as being those of extreme left lower extremity fatigue, discomfort on the inside of her left foot, as well as a "swing in my left leg". The initial onset of these symptoms was noticed with running. The patient sought out extensive medical care for these symptoms over the years, stating that no specific diagnosis was made until November 2006. At that time, exercise induced compartment syndrome was diagnosed, and she underwent a left lateral compartment release on March 13th, 2007. Her symptoms resolved following the surgery, with the exception of persistent left medial forefoot discomfort with any amount of walking. She also states that she feels as if her left foot "twists toward the inside" and this causes her to catch her great toe, which then causes her to stumble. Patient has not been able to return to running. MRI and X-ray findings of her left foot and ankle are reported by the patient as being "normal". Orthotic management has been attempted but unsuccessful in reducing her symptoms. Patient reports being "active", but not involved in a formal physical therapy program. Patient goals would be to be able to walk without discomfort and resume a running program.

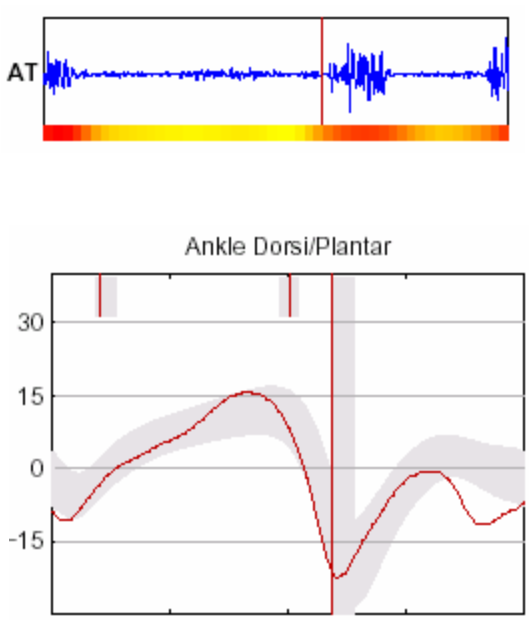
Clinical Data:

Patient rates her walking level at 10 on the Functional Assessment Questionnaire walking scale. She reports being able to perform 20 of 21 functional skills, with the skill of running marked as "very hard". Close up videos suggest evidence of mildly excessive ankle joint plantarflexion in late swing on the left, with recruitment of the extensor hallicus longus throughout the swing phase of the gait cycle, more noticeable on the left than on the right. Plantar pressure data demonstrates high areas of pressure throughout the heel and forefoot bilaterally, consistent with the high arched architecture of her feet bilaterally. Asymmetries are evident on plantar pressure data, however, which show decreased pressure under the great toe, and increased pressure under the 1st metatarsal head of the left foot as compared to the right. Physical exam findings were unremarkable, with anterior tibialis strength graded at 5/5 with manual muscle testing.

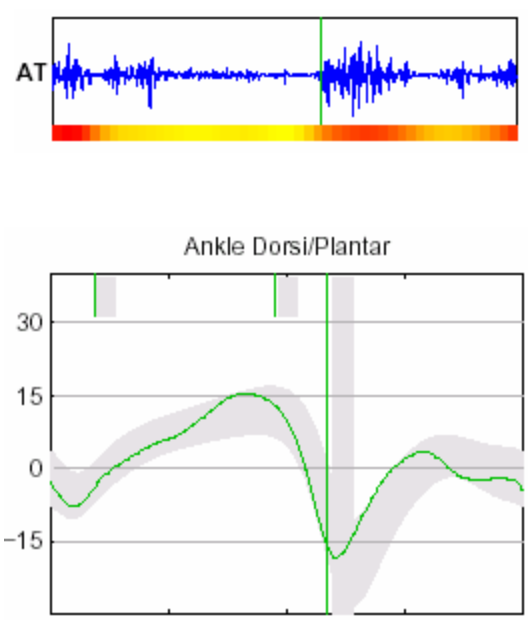
Gait Data:

At the ankle, typical kinematic and kinetic data is noted during stance phase of gait bilaterally. Swing phase kinematics show evidence for excessive ankle joint plantarflexion in mid-swing on the left only. This is accentuated with running. Dynamic swing phase EMG demonstrates a slight difference when comparing the left to the right sides. A fairly typical signal for the anterior tibialis was seen bilaterally i.e. early activation followed by a period of relative quiet in mid-swing followed by a late burst of activity. However, the onset of the second burst of activity is delayed and somewhat diminished on the left as compared to the right. This coincides with the time she loses

ankle dorsiflexion during swing on her left side, therefore, predisposing her to catching her toe.



Left sagittal plane AJ kinematics/AT EMG



Right sagittal plane AJ kinematics/AT EMG

Treatment Decisions:
Clearly pes cavus foot positions are present bilaterally and have been so throughout the patient's lifetime. These may have some slight contributory affects to her pain pattern, but we believe that the catching of her toe and the left foot pain pattern are clearly related to the mild dysfunction of her left anterior tibialis, and the secondary compensation with use of the extensor hallicus longus (EHL). Therefore, treatment recommendations were made and directed toward this. Options would include:

1. Physical therapy for tibialis anterior strengthening
2. Consideration of an electrical stimulation program for the anterior tibialis either for therapy of from a functional standpoint (Bioness orthosis)
3. Use of an ankle-foot orthosis (AFO) as needed to control symptoms

Summary:
In this case, use of gait analysis allowed us to see subtle, but relevant, dysfunction in this patient's anterior tibialis muscle. This dysfunction had not previously been detected with use of gait by observation and physical exam alone. Through correct identification of the problem, treatment was recommend and, if carried through with, will hopefully result in the patient achieving her goals of pain free ambulation and the ability to resume a running program. The patient was referred back to her referring orthopedist, and as of yet, has not had a follow-up gait study.

Gait Analysis with Dynamic EMG for Decision-Making in an Adult with a Foot Drop of Unknown Etiology: A Case Study

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Patient History

The patient is a 68-year-old woman with a right foot dysfunction of unknown etiology. For years she had walked on a treadmill daily for exercise. Two years prior to testing she noticed her foot was dragging initially during her workouts and eventually throughout the day. She had difficulty walking due to excess plantar flexion during swing limb advancement that caused her to trip (and sometimes fall). She had increased her workouts and had some low back pain but otherwise nothing had changed. She had seen numerous orthopedists and neurologists for treatment but still did not have a clear diagnosis other than "drop foot". An MRI of her low back showed DJD and an MRI of her brain was negative. A diagnostic EMG and a nerve conduction study of her lower leg also were not significant. Previously she tried a dorsiflexion assist brace, exercise and treadmill training (in PT) that she felt did not help her. She was referred for evaluation of her muscle timing during gait to assist in treatment planning.

Clinical Data

Relevant clinical findings included limited passive range of motion into dorsiflexion with the knee extended and flexed (-10 degrees), and hip extension (-15 degrees). Strength testing revealed moderate weakness in all muscles of the right hip: 4-/5 hip extensors and 4/5 hip abductors by manual muscle testing. Muscle strength in the quadriceps and all muscles of the foot and ankle were tested at 5/5 strength. Tone was normal throughout the extremity.

Gait Data

Her free walking velocity was moderately decreased at 47 meters/min (64% of normal) due to a decrease in both stride length and cadence. Footswitch recording indicated that initial contact on the right was made with the medial forefoot and heel off was markedly delayed at toe-off.

Quantitated motion analysis of her right lower extremity during barefoot walking identified primary gait deviations including equinus throughout swing, increasing markedly in mid and terminal swing. In stance her ankle rapidly progressed to excessive dorsiflexion reaching 17 degrees in terminal stance. (Figure 1) Her rearfoot was inverted from terminal swing through mid stance. At the hip and knee, extension in swing also was premature. In stance, hip and knee extension were inadequate with 15 degrees of residual flexion.

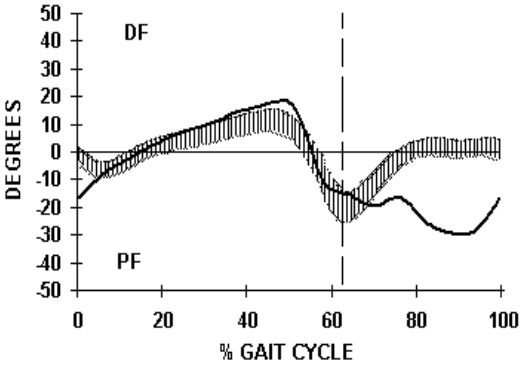

Figure 1: Sagittal ankle plane kinematics

Dynamic EMG recording during walking identified markedly disrupted timing of muscle activity consistent with impaired motor control. Soleus (SOL) had appropriate activity in mid and terminal stance, but had a second burst of vigorous activity during initial and mid

swing. Gastrocnemius (GAST) had low-level activity in both stance and swing. Of the anterior compartment muscles, anterior tibialis (AT) and extensor digitorum longus (EDL) had delayed onset with dominant activity from late mid swing through mid stance. All of the other perimalleolar muscles including posterior tibialis (PT), peroneus longus (PL), peroneus brevis (PB), and flexor digitorum longus (FDL) had low level activity in stance and a second more vigorous burst in mid and terminal swing with particularly strong swing phase activity in PT and PL. (Figure 2)

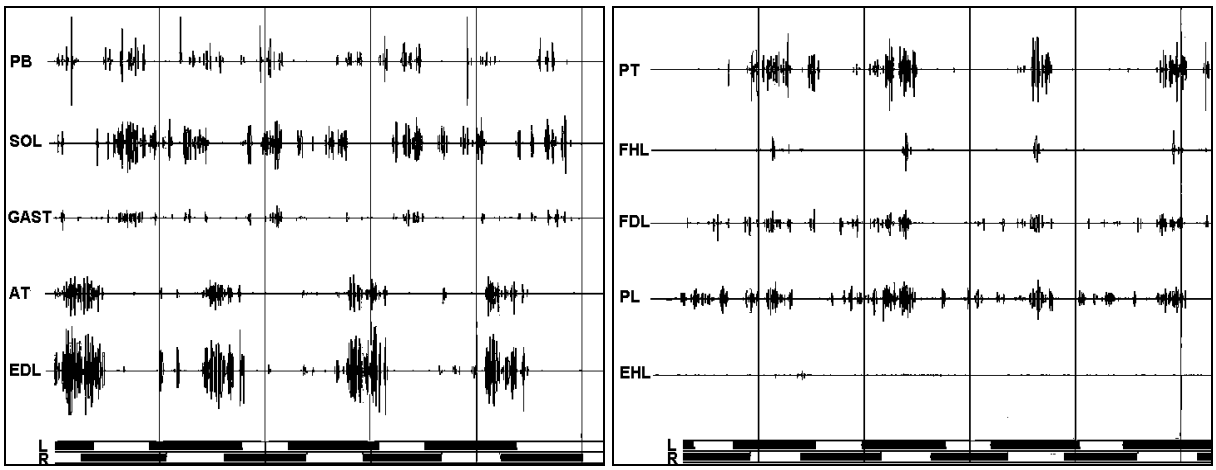

Figure 2: Dynamic EMG of the right lower extremity during free walking

Treatment Decisions and Indications

Her plantarflexion in swing is a consequence of impaired motor control with abnormal muscle phasing. This suggests a central origin rather than peripheral and would suggest impairment in areas other than cortical possibly basal ganglia. This pattern is consistent with focal dystonia.

Her volitional, selective control is less impaired than her automatic control during gait. The trigger of her inappropriate activity appears to be related to periods of stretch or lengthening of the plantarflexors. Soleus is tight and was more vigorously active than gastrocnemius. The best option to control her active plantarflexion in swing would be to alter her muscle activation pattern. This is difficult to do, but would be worth a trial of an electrical stimulator unit (Bioness or WalkAide) to the anterior compartment during gait to see if the anterior compartment muscles could be facilitated in swing and the posterior compartment muscles inhibited. It is also recommended that she perform stretching exercises to reduce her ankle plantar flexion contracture.

If the electrical stimulation and stretching does not improve her plantarflexion in swing, she could use an articulating ankle foot orthosis to control her ankle in swing. Her plantar flexion is strong enough that she would need a plantar flexion stop in addition to dorsiflexion assistance. Additionally, temporary blocks to the soleus and then peroneus longus muscles could help determine the muscular contributions to her plantarflexion in swing. Botox injections or a transfer of peroneus longus to the dorsum of the foot could be considered if the plantar flexion was reduced following the temporary blocks.

Summary

The gait analysis study with EMG activity allowed us to identify inappropriate muscle activity and make appropriate recommendations.

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Gait Analysis to determine changes in gait pattern due to fatigue in a child with cerebral palsy spastic diplegia

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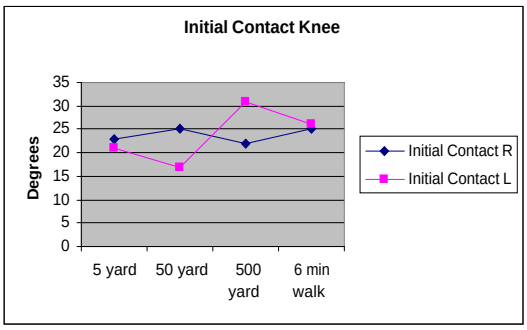
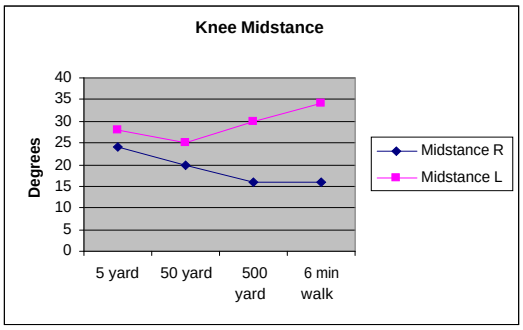
Patient History: The patient is a 9 year old girl with cerebral palsy, spastic diplegia. GMFCS Level II. After gait analysis, she underwent single event multi-level surgery in August 2007 including bilateral hamstring lengthenings, bilateral rectus femoris transfers and left percutaneous heel-cord lengthening. Post operatively she received PT 2 x a week and was prescribed and wears bilateral solid ankle foot orthoses.

Clinical Data: Physical exam (4/08) included the following:
Mild increased tone in the LE's L>R. (1+ Modified Ashworth Scale)
Popliteal Angles: 80°R, 70°L
Ducan Ely test: Negative Bilaterally
Ankle Dorsiflexion: 10°R, 5°L (knee extended)
Minimal extensor (quadriceps) lag 5°R, 10°L
Minimal persistent crouch L > R
Decreased one leg standing balance Left

Gait Data: Kinematic and kinetic data was collected at each distance of the Functional Mobility Scale (5 yards, 50 yards, 500 yards)[1] and at the end of a 6 minute walk test. Heart rate was monitored at baseline prior to walking and at the end of each distance. Rest was given between each distance interval to allow hear rate to return to baseline prior to beginning the next distance. Energy Expenditure Index (EEI) was calculated as were temporal-spatial gait parameters.
With increased distance walked, the following was observed:

- Increased knee flexion at initial contact on L at 500 yards and 6 minute walk test
- Progressively increasing knee flexion in mid-stance on Left
- Decreased Cadence
- Decreased walking speed L>R
- Increased double limb support time on Right
- Significant increase in EEI





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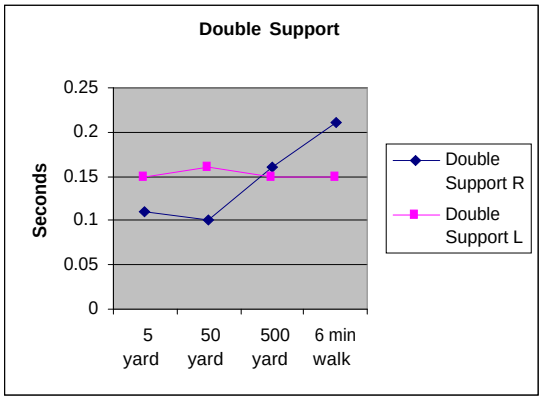

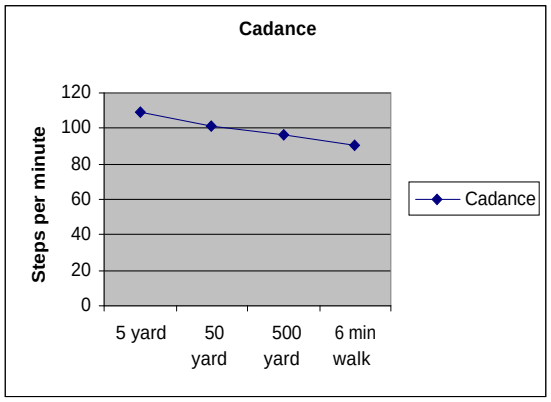

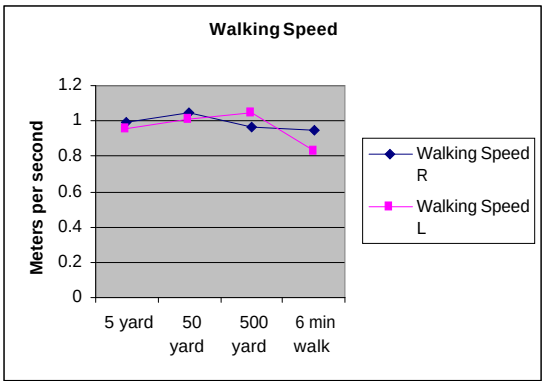

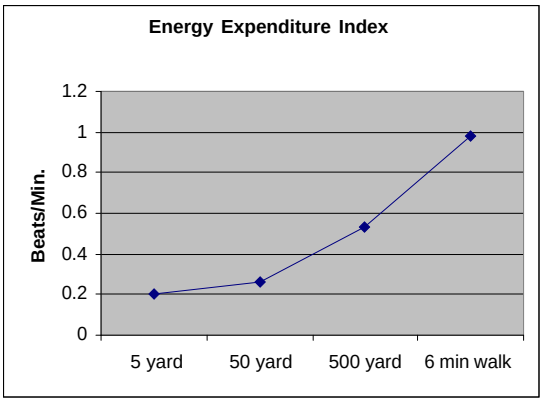

Treatment Decisions and Indications:

The results of this study indicate that as this child walked further and became fatigued the function of the left LE declined more than the right. Increased knee flexion in mid-stance and increased knee flexion on initial contact may reflect greater quadriceps weakness on the left side. Double limb support time on the right increased as the child relied on the stronger leg for support and stability. Walking speed on left side is also decreased more than right. The changes suggest that physical therapy interventions should be directed not only at strengthening LE musculature but strengthening for endurance and should be targeted to the appropriate areas. Additionally, the rapid increase in EEI suggests that training for cardiovascular fitness is also recommended.

Summary:

Gait analysis was useful in this case in documenting changes in gait pattern seen due to fatigue. The decline in gait parameters seen at the relatively short distances of the FMS was greater than expected. When assessing children with CP the gait laboratory the results may not reflect the gait pattern used when walking in the school or community environment where greater distances are walked and potential for fatigue greater.

1. Functional Mobility Scale (FMS) Graham HK, et al. *J Pediatr Orthop* 2004;24:5 514-520.

Case Study: MH Pre and post Single Event Multilevel Surgery – An illustration of the use of the Movement Analysis Profile to monitor progress

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[2]Gait CCRE, Murdoch Childrens Research Institute, [3]University of Melbourne

Patient History: MH a boy with spastic diplegia, GMFCS level II, Functional Mobility Scale (FMS) 5, 2, 5 referred to the gait laboratory for the evaluation of gait and function prior to multilevel surgery at age 7 years 8 months. His parents stated concerns were intoed gait, twisted legs, inability to wear AFOs and deteriorating function with increasing falls at school, requiring use of a posterior walker for safety reasons.

Clinical Data: The principal findings on physical examination were:

- Increased internal rotation, decreased external rotation at both hips, left worse.
- Increased femoral neck anteversion, bilaterally.
- Reduced left hip abduction
- Bilateral equinus contractures, left worse, affecting mainly gastrocnemius
- Reduced popliteal angles and hamstring tightness bilaterally, right worse.
- Hip extensor and hip abductor weakness
- Height 119.2 cm, weight 20.8 kg.
- GMFM 66 - 65%

Gait Data: Gait data are summarised using the Movement Analysis Profile (MAP)[1] in Figure 1.

Treatment Decisions and Indications: A multilevel surgery program was recommended but there was concern regarding MHs weight and he was referred to a developmental pediatrician to assess his nutritional status prior to surgery. MH had a further gait laboratory evaluation 12 months later. His parents were concerned about further deterioration in function with MH using a posterior walker for the majority of the time. His FMS score had deteriorated to 5, 2, 2. GMFM 66 was unchanged. Further physical examination revealed the following problems:

- Fixed flexion deformity at both hips
- A further increase in internal rotation and reduction in external rotation at both hips
- A further reduction in the range of hip abduction bilaterally
- Height 123.4 cm, Weight 22.85 kg.

A substantial deterioration in kinematics across all planes affecting the left lower limb was seen as is indicated by the increased variable scores on the MAP in Figure 1 (b).

The surgical prescription following the second evaluation, and based on full kinematic and kinetic data, was as follows:

- Bilateral proximal femoral varus derotation osteotomies left more than right
- Bilateral psoas lengthening over the brim of the pelvis
- Bilateral percutaneous lengthening of adductor longus
- Left medial hamstring lengthening with transfer of semitendinosus to the adductor tubercle

- Left gastrocnemius lengthening plus soleal fascial lengthening and intramuscular lengthening of tibialis posterior
- Right gastrocnemius lengthening

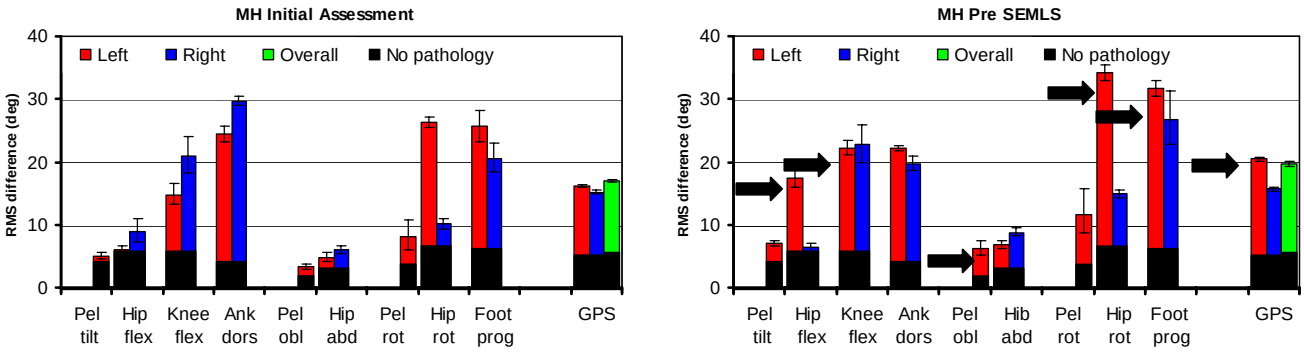

Figure 1 Movement Analysis Profile at initial assessment (left) and pre Single Event Multilevel Surgery (SEMLS) (right). The arrows indicate areas of most deterioration.

MH was reviewed in the gait laboratory at 12 months post surgery. His gait pattern (Figure 2(a)) was noted to have improved considerably but a mild persistent knee flexion deformity was noted clinically on the right side with persistent hip and knee flexion in the sagittal plane. A right medial hamstring lengthening and transfer of semitendinosus to the adductor tubercle was carried out at the time of blade plate removal. This resulted in a further improvement in gait parameters at 24 months after multilevel surgery (Figure 2(b)). Mobility and function showed improvement at 24 months with an FMS of 5, 5, 5 and an improvement of 6% on the GMFM 66. At this time knee flexion persists during stance and monitoring will need to continue with the potential for growth plate surgery prior to puberty if required.

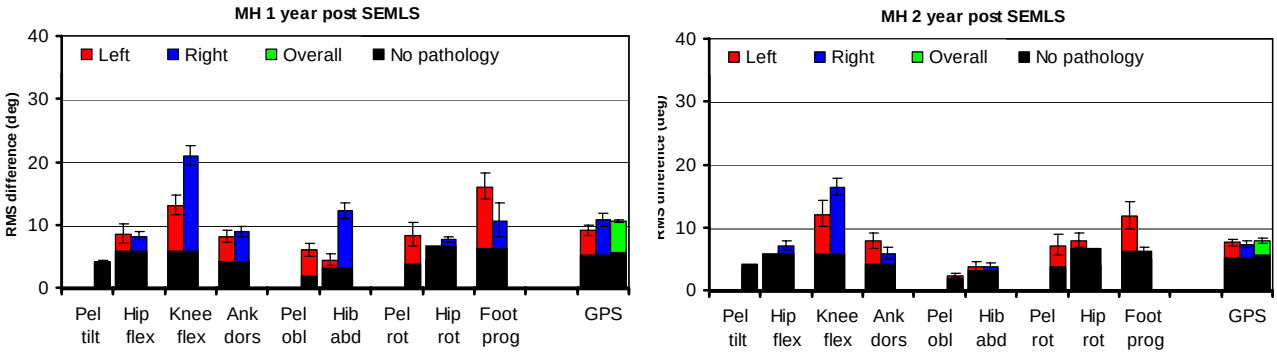

Figure 2 Movement Analysis Profile at 12 (left) and 24 months (right) post Single Event Multilevel Surgery (SEMLS)

Summary: This case illustrates how instrumented gait analysis is used in the decision making process and follow up assessment for children undergoing SEMLS. The use of the MAP highlights the magnitude of deviations from gait without pathology as well as changes over time.

Reference: [1] Baker, et al Gait & Posture 28S (2008) S7

Gait Analysis for Treatment Decision-making in Persons with Cerebral Palsy: A Case Study with a Focus on Transverse Plane Joint Kinematics

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PATIENT HISTORY

The patient is a 5+8 year old boy with a diagnosis of cerebral palsy (CP), spastic diplegia. He was the product of a premature birth at 28 weeks gestation with a birth weight of 2lbs. 7oz. He first began walking with Lofstrand crutches at age 4. He has had no previous lower extremity surgery, but has had BOTOX injections to bilateral medial hamstrings and bilateral gastrocnemius muscles. He is currently taking oral Baclofen to manage his spasticity. He wears bilateral solid ankle foot orthoses (AFO). His gross motor function classification system (GMFCS) level is III and he ambulates primarily with bilateral Lofstrand crutches. He was referred for gait analysis for orthopedic decision making due to increased falls and to better clarify his rotational profile.

CLINICAL DATA

Passive internal rotation at bilateral hips was 70º, passive external rotation was 45º right, 40º left and palpated femoral anteversion was 60º right and 55º left. Foot thigh angle was 0º right and -10º left and bi-malleolar axis was 10º right and 5º left. Popliteal angles were -60º bilaterally. Passive ankle dorsiflexion was -15º right and 0º left (knee extended) and 0º right and 15º left, (knee flexed). Abduction was 25º right and 10º left (hip at 0º) and 40º bilaterally with the hips at 90º. Isolated motor control was poor throughout both lower extremities.

GAIT DATA

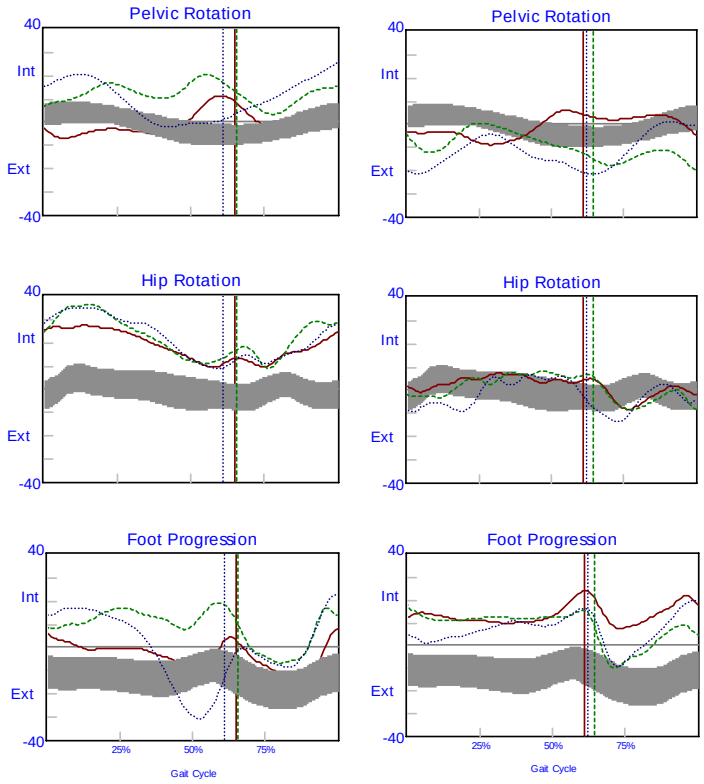

Figure 1: Transverse plane profile for 3 gait cycles on both the right (far left) and left sides (next left) (Typical reference = gray band)

A visual evaluation of the patient's gait showed a toe walking gait pattern with crouch and limited sagittal plane range of motion. In the transverse plane, foot progression varied from internal to external and the degree of inward pointing of the knees also varied.

The transverse plane kinematic profile for the pelvis, hip and foot progression during barefoot walking with Loftstrand crutches (Fig. 1) showed the following:

1) Variable degrees of internal rotation of the right hemi pelvis and external rotation of the left hemi pelvis
2) Consistent internal rotation of the right hip and neutral rotation of the left hip
3) Variable foot progression of the right foot ranging from neutral to internal and variable but consistently internal foot progression of the left foot

TREATMENT DECISIONS AND INDICATIONS

1) A right femoral derotation osteotomy: supported by clinical examination results of increased femoral anteversion and consistent internal hip rotation during gait
2) No left femoral derotation osteotomy: although clinical examination showed increased femoral anteversion, the gait data showed consistent neutral hip rotation during gait.
3) A left tibial derotation osteotomy: supported by increased internal tibial torsion noted on clinical assessment and consistently internal foot progression in the presence of neutral hip rotation and external pelvic rotation.
4) No right tibial derotation osteotomy: supported by less and milder tibial torsion on clinical assessment in comparison to left side and internal foot progression that will change post surgery due to femoral derotation on same side
5) Soft tissue procedures were also recommended to address the bilateral toe walking and crouch/stiff knee gait pattern and included bilateral lengthening procedures to the following muscles or muscle groups: medial hamstring, intramuscular rectus femoris, hip adductor and gastrocnemius.

SUMMARY

The transverse plane kinematic data provided objective data that assisted in determining the primary issues related to transverse plane abnormalities for this patient during walking. The motion data confirmed the primary source of stride to stride variability which was related to pelvic motion variation stride to stride and on the right side, variable clearance issues. The motion data confirmed very consistent and asymmetric hip rotation with increased internal rotation of the right side and typical rotation on the left despite femoral anteversion noted on clinical assessment, bilaterally. The basis for the unilateral surgical decision at the hip was related to the asymmetric gait analysis data. The internal foot progression on the left was secondary to internal tibial torsion as noted on clinical assessment. The variable foot progression on the right was due in part to variability noted at the pelvis and variable toe drag at toe off. On both sides, the degree of internal torsion was confounded by a bilateral mid foot break. The unilateral left tibial derotation osteotomy was secondary to the increased internal tibial torsion on the left side and consistent internal foot progression on the left.

In this case, the transverse plane motion analysis data helped to provide clarification for an asymmetric transverse plane profile noted on both the clinical assessment and visually. The motion data also provided confidence that the femoral anteversion was a consistent problem and that variation in degree of in-pointing knee was related to pelvic variability in the transverse plane and not hip rotation variations. Therefore, the gait data provided clarification about where variable motion was present which impacted the surgical decision.

Resistance Training Alters Joint Moments and Powers in Multiple Sclerosis Patients Independent of Disease Severity

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Introduction

Multiple Sclerosis (MS) is a progressive neurological disease which affects over 400,000 Americans. Symptoms vary widely among patients and may include sensory disturbances, optic neuropathy, trunk and limb parasthesias, limb weakness, gait ataxia, and severe fatigue[1]. Treatment options may include medication, physical therapy, and even steroids. Physical activity had previously been discouraged in order to protect the current physical functioning of the patient. Recently, exercise interventions have been shown to positively affect MS patients with respect to aerobic fitness and quality of life measures[2]. The purpose of this study was to determine the effects of a progressive resistance training program on the lower extremity joint moments and powers during walking in MS patients. It was hypothesized this training would alter both joint moments and powers in MS patients. It was also hypothesized that MS patients with more severe symptoms would have greater changes in the gait parameters than less severely affected MS patients.

Clinical Significance

FDA approved pharmacologic therapy has effectively decreased the number of MS relapses and their severity in the relapsing-remitting MS population. However, MS patients continue to experience disease progression, fatigue, muscle weakness and balance problems either due directly to the disease[3], progression of established disability, or as a side effect of the immunotherapy[4]. Physical rehabilitation has been shown to have a positive effect in MS patients[5] in improving physical functioning. However, inpatient rehabilitation programs may not be routinely available or covered by insurance. Evaluation of resistance training as a rehabilitation tool, which could be readily available to MS patients, could create a viable treatment option for MS patients to improve walking capability.

Methods

Twelve MS patients (age 43.7 ± 10.9 yrs; EDSS 3.9 ± 1.6) walked through a 10 meter walkway at their self-selected pace, while kinetics and kinematics were collected for 10 trials with a Kistler force plate (600Hz) and an 8-camera Motion Analysis system (60 Hz). Data collection was performed at baseline (C1), after 3 months (C2), and after 6 months (C3) of participation in a twice weekly supervised, progressive resistance training program that included lower extremity, upper extremity, and core exercises. Lower extremity joint moments and powers were calculated from the kinetics and the kinematics. The severity of disease for each patient was assessed according to the Kurtzke Disability Scale[6] and each patient was assigned an EDSS score. Patients were then grouped according to this score: Group 1 with no clinically defined mobility problems had an average EDSS of 2.7 ± 1.1,

Group 2 with minimal to moderate mobility and balance problems had an average EDSS of 5.3 ± 0.8. A 3x2 (testing time by groups) mixed ANOVA was performed on the group means of the identified peaks of joint moments and powers during stance.

Results

No group main effects were identified for joint moments. For joint powers, ankle power generation during late stance was significantly decreased for the most severe MS group. Regarding the testing time main effects, knee extension moment was significantly increased from C1 to C3 ($p = .003$). Knee power generation during early stance was significantly increased from C1 to C2 ($p = .007$) and knee power absorption during late stance was also significantly increased from C1 to C2 ($p = .014$) and from C1 to C3 ($p = .000$). Hip power absorption during midstance was significantly increased from C1 to C2 ($p = .012$), while hip power generation during late stance was significantly increased from C1 to C2 ($p = .040$) and from C1 to C3 ($p = .015$). There were no significant differences for any of the joint moment or power variables between C2 and C3. No significant interactions were also found.

Discussion

The resistance training intervention resulted in significant changes in joint moments and powers at the knee and hip. All changes identified were increases in joint power generation or absorption which would result in improved ability in the patient to transfer weight during stance. Though the training protocol for this study required a full 6 months of resistance training, the results may also indicate that just three months of training can induce sufficient lower extremity changes that would translate to improved gait.

Only one significant difference was present due to disease severity, and because no significant interactions were found between severity and testing time, this difference in ankle joint powers remained even after training. The lack of significant differences between disease severity groups may indicate that gait analysis parameters may not support EDSS as the most efficient tool to distinguish MS patients according to severity.

Future research work will focus in generalizing these findings with comparisons with healthy matched controls and a larger sample size. In addition, the exact duration of resistance training, required to improve gait and alleviate symptoms in MS patients, needs additional research, as well as to identify retention gains or losses.

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Effects of Locomotor Treadmill Training on Gait Symmetry in Acute Stroke

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INTRODUCTION

Gait impairment is a common result of stroke. There is little consensus as to the optimal treatment in order to reestablish normal gait after stroke. Studies have shown that it is difficult to change a gait pattern once established.[1,2] The potential to restore gait without the typical asymmetries would represent significant progress in the rehabilitation of individuals after stroke. The purpose was to investigate the results of locomotor treadmill training with partial body weight support (BWS) on gait symmetry prior to over ground gait in acute stroke.

CLINICAL SIGNIFICANCE

A new therapy regiment has shown potential to improve gait after stroke. Returning post-stroke patients to a more normal and efficient gait pattern may create a new, better standard of care for stroke patients.

METHODS

Seven subjects who received traditional inpatient therapy (TIT) and 7 subjects that received locomotor treadmill training (LTT) with partial body weight support were recruited. Kinematic data were recorded with a 12-camera Vicon motion capture system. Patients were tested during level walking at least 6 months post stroke. Gait symmetry parameters included single limb support and sagittal plane angles of the hip, knee, and ankle.

Treadmill training was initially set at 30% BWS and speed at 0.7 mph. The goal in subsequent sessions was to increase BWS by 5% and speed of 0.1 mph. Treadmill training was discontinued when participants were able to walk 10 continuous minutes at a speed of 1.5 mph without rest and without any support. All subjects achieved full weight-bearing on the treadmill. Two trained staff members assisted the subjects to facilitate normal loading, stance, and swing components for the affected lower extremity and to facilitate normal weight shift at the pelvis during each training session.

RESULTS

Single limb support times were significantly different, with the treadmill group more symmetrical. Hip flexion at initial contact between groups was significant, with less variation between limbs in the treadmill group. Maximum knee flexion differences during swing between groups did reach statistical significance, as did maximum knee extension during stance. Lastly, there was no difference between limbs for maximum ankle dorsiflexion at terminal stance between groups.

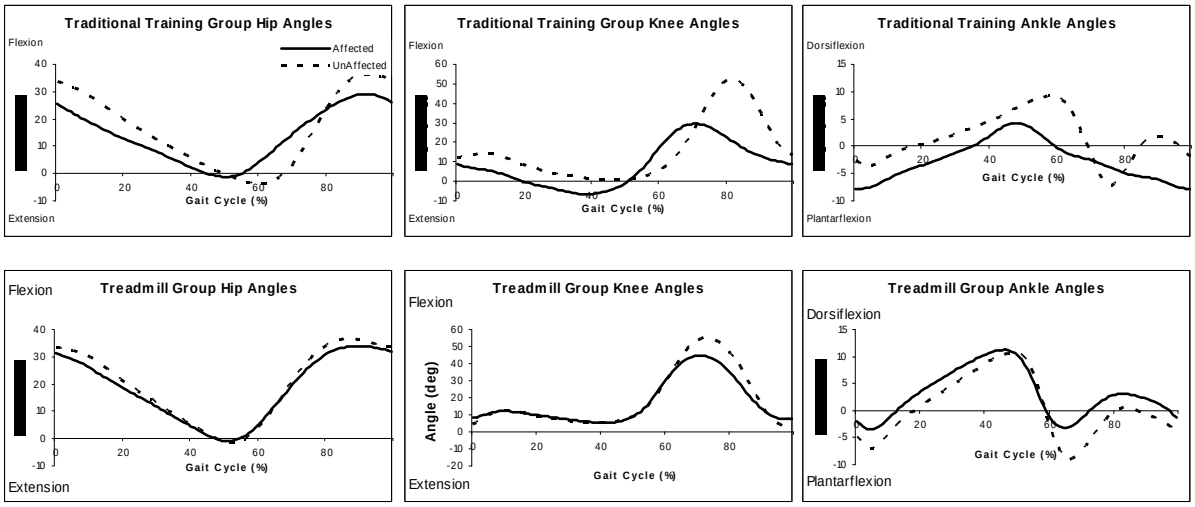

Figure 1 Mean sagittal plane angles of affected and unaffected sides for both locomotor treadmill training group and traditional inpatient therapy group

DISCUSSION

Gait after stroke is often slow, asymmetrical and inefficient, impacting the individual's ability to resume meaningful roles in society. Participants post stroke that initiated gait training on the treadmill with partial BWS prior to the initiation of over ground gait demonstrated better gait kinematics, symmetry, and velocity than participants who received traditional gait rehabilitation

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THE EFFECT OF BODY WEIGHT SUPPORT TREADMILL TRAINING ON PARETIC LEG CONTRIBUITION IN HEMIPARETIC WALKING IN PERSONS WITH CHRONIC STROKE

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INTRODUCTION

Post-stroke hemiparesis is one of the most debilitating impairments following stroke due to its effects on locomotion. The gait of individuals with post-stroke hemiparesis is characterized by reduced walking speed [1], uncoordinated movement patterns [2], and decreased force output by the paretic limb [3]. Body weight support treadmill training (BWSTT) is a task-specific therapy approach used for gait re-education in individuals with post-stroke hemiparesis. This gait-training strategy implements the use of an overhead suspension and harness system to support a percentage of a patient's body weight, symmetrically unloading weight from the lower extremities. The focus of BWSTT is repetitive forced use of the paretic limb while continuously practicing complete gait cycles without the risk of falling. It has been suggested that gait re-education with BWSTT leads to a more successful recovery of ambulation with respect to overground walking speed and endurance [4, 5]. However, despite improvements in overall gait speed, the effects of BWSTT on the paretic limb are not well understood in individuals with post-stroke hemiparesis. The purpose of this study was to examine the effect of BWSTT on walking velocity, paretic leg propulsion, and the mechanical work produced at the hip, knee, and ankle of the paretic leg.

CLINICAL SIGNIFICANCE

The positive changes in walking speed and endurance following BWSTT may be the result of more effective compensatory mechanisms by the nonparetic limb, rather than improvements in the paretic limb's function. Understanding the effects of BWSTT on paretic limb function during gait may provide more insight into the efficacy of BWSTT as a viable treatment option for individuals with post-stroke hemiparesis.

METHODS

Thirteen individuals with chronic stroke (11 female; 2 male; 60.4 ± 11.2 years of age; time since stroke (years) = 3.4 ± 2.7; affected side left = 3, right = 10) participated in the study. Written informed consent, approved by the Ball State University Institutional Review Board, was obtained from all participants. Each participant was required to complete 24 sessions of BWSTT over 8 weeks, with each session consisting of 20 minutes total walking time. A comfortable 10-meter walk test (CWT), used to assess overground gait velocity, and a three-dimensional gait analysis were performed on all participants prior to and following the intervention period. Bilateral kinematics and kinetics were captured at 60Hz with a 10 camera VICON F-series motion capture system and processed with the Plug-in-Gait model

(Vicon, Lake Forest, CA, USA). The percentage contribution of the paretic limb to total propulsive impulse was calculated using methods similar to those used by Bowden et al [6]. Additionally, joint moment work at the hip, knee, and ankle of the paretic and nonparetic limbs were calculated. Paired t-tests were used to analyze the differences in mean values of gait velocity and percentage contribution of the paretic limb to total propulsive impulse. A repeated measures ANOVA was applied to analyze the differences in mean values of the relative contribution of the paretic hip, knee, and ankle to the total positive work.

RESULTS

Overground gait velocity was significantly increased following the 8-week BWSTT intervention program. However, there were no significant differences found following BWSTT for the percentage of total propulsive impulse generated by the paretic limb or the percent contribution to total positive work of the paretic hip, knee, or ankle. The means and standard deviations for all comparisons are presented in Table 1.

Table 1: Means and SD for all Dependent Variables

	Pre-Intervention	**Post-Intervention**	**p-value**
CWT Walking Velocity (m/s)	0.62 ± 0.14	0.73 ± 0.17	.003*
% Total Propulsive Impulse - Paretic Limb	34 ± 19	34 ± 17	.918
% Contribution to Total Positive Work			
Paretic Hip	43 ± 13	42 ± 13	.821
Paretic Knee	27 ± 15	27 ± 18	.930
Paretic Ankle	30 ± 15	31 ± 14	.919

*indicates significant difference at alpha level of 0.05

SUMMARY/CONCLUSIONS

Similar to previous results [4, 5], overground walking velocity significantly increased following BWSTT. However, the contribution of the paretic limb to hemiparetic walking, with respect to its relative contribution to total propulsive impulse and total positive work, remained unchanged following the intervention. These results suggest that the subjects were achieving faster walking velocities by some mechanism other than changes in paretic leg contribution. Further research is needed to address and identify the mechanisms that are contributing to the increased walking velocities.

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Surface EMG Does Not Accurately Reflect Rectus Femoris Activity Throughout the Gait Cycle: The Influence of Vasti Crosstalk in Multiple Gait Conditions

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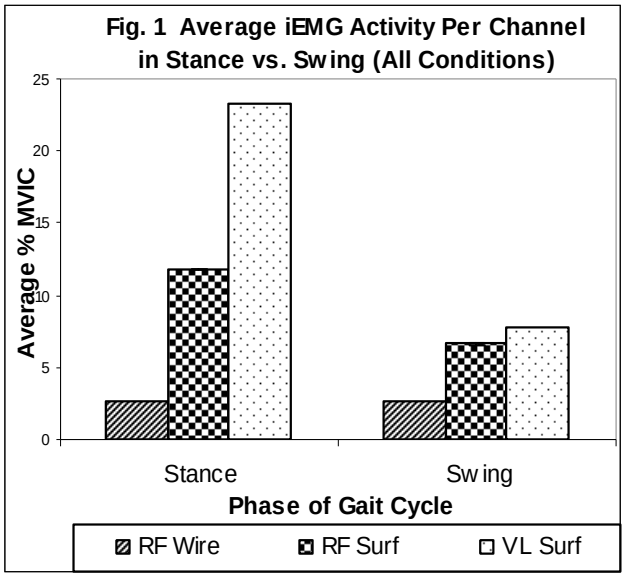
Introduction: Surface electromyography (sEMG) of the rectus femoris (RF) has been commonly described in the literature for recording rectus femoris (RF) activity during gait[2,3,5,7]. This technique for monitoring RF continues to be popular in clinical gait analysis laboratories. Yet multiple authors have described RF sEMG being susceptible to crosstalk from the neighboring vasti[1,4,6]. These reports have generally consisted of small sample sizes, limited (or non-gait[1]) conditions, and/or descriptive reporting of EMG findings. Therefore, the purpose of this study was to quantitatively analyze the impact of vasti crosstalk on surface-recorded RF EMG during both the stance and swing phases of gait over a range of speed and crouch conditions in a larger sample of able-bodied adults.

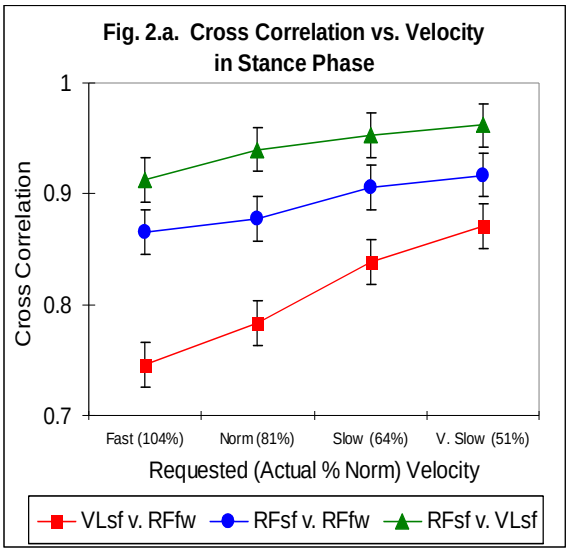
Clinical Significance: Vasti crosstalk in RF sEMG data can lead to misinterpretation of RF activation during gait and thus affect related clinical recommendations.

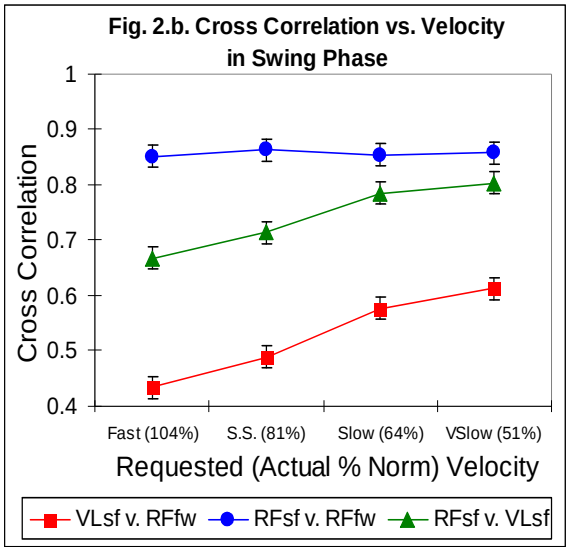
Methods: Nineteen able-bodied adults (ages 18-40) were tested walking in 16 different combinations of speed (4) and crouch (4) in attempt to create a broad range of vasti and RF activation patterns across the gait cycle. RF was monitored using both a set of paired fine wire electrodes (RFfw) and a surface electrode (RFsf), while vastus lateralis (VL) was monitored using only sEMG (VLsf). Signals were low and high pass filtered, rectified, low pass filtered (to generate linear envelope), normalized to maximal voluntary isometric contraction (MVIC), then integrated (iEMG) to find the average percent maximum voluntary isometric contraction (%MVIC) for each muscle in both the stance and swing phases as measured by each of the 3 electrodes (RFfw, RFsf, VLsf). Cross correlations of the time series data between each pair of channels, as well as the average %MVIC (from iEMG data) as measured by each of the 3 electrodes (RFfw, RFsf, VL sf), were compared for all 3 electrode channels during both the stance and swing phases of gait. All data were analyzed using ANOVA's with statistical significance set at $p < 0.05$.

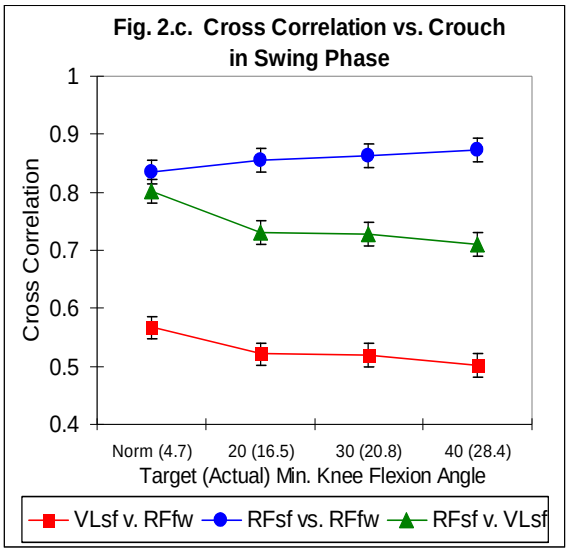
Results: As shown in Fig. 1, on average across all conditions, VLsf demonstrated roughly 3 times more activity in stance than swing phase, while RFfw showed equal magnitude of activity in both phases. In contrast to RFfw, RFsf suggested that RF activity was higher in stance than swing. Indeed, during stance, RFsf correlated better with VLsf than it did with RFfw (Fig. 2a). In swing, RFsf appropriately correlated best with RFfw (Figs. 2b and 2c).



Additionally, in both stance and swing, significant interaction effects ($p<0.0001$) were found in the cross-correlation data, where faster walking resulted in increasingly different activation patterns for VLsf and RFfw (Figs. 2a and 2b). The RFsf and RFfw correlation remained high for all speed conditions during







swing (Fig. 2b). A significant interaction effect was also found for crouch during swing ($p<0.0001$), where, with increasing crouch, the two RF channels became slightly more similar, while the correlations indicate VLsf and the RF channels were less similar (Fig. 2c.).

Discussion:

The marked increase in the magnitude of vasti activity during stance leads vasti crosstalk to taint RF sEMG during stance phase. Cross-correlation data for stance quantifies the inconsistency between actual RF activity (by fine wire) and VL: Consistently higher correlations between RFsf and VLsf across all walking speeds in stance phase is problematic for the accurate interpretation of RF sEMG.

Cross-correlations in swing phase also demonstrated divergence between RFfw and VLsf activity. With increased speed, VL activity is likely increased in late swing in preparation for weight acceptance, activity which may also be picked up by RFsf. However, true RF activity is only known to increase with speed during swing initiation. This puts the curves more out-of-phase (and explains the lower cross-correlations). In swing phase, unlike stance, for all conditions, RFsf and RFfw were the 2 channels whose activation patterns most highly correlated.

Apparent RF sEMG activity observed during the stance phase of gait should be regarded with considerable skepticism, and even more so in conditions where higher levels of vasti activation might be expected. Gait laboratories should be cautious when scaling RF sEMG data to accommodate the amplitude of all apparent RF activity during stance, as this likely includes vasti crosstalk and can lead to under estimation of actual, lower amplitude RF activity during swing. The accurate recording of RF activity throughout the gait cycle across a variety of gait patterns necessitates the use of fwEMG.

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ELECTROMYOGRAPHY IS BETTER IN QUANTIFYING MUSCULAR DEMANDS THAN JOINT MOMENTS WHEN CO-CONTRACTION EXISTS

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INTRODUCTION

The ratios of the peak joint moments during activities of daily living to the peak torques measured during maximum voluntary contraction (MVC) (N-moment; Hortobagyi et al. 2003) and the ratios between the EMG amplitudes collected during activities and during the MVC (N-EMG; Hahn et al. 2005) were used to quantify the muscular demand during gait. Lee (2005) demonstrated that both ratios showed a similar ability in detecting functional challenge applied onto lower extremity muscles, and implicated that using the N-moment ratio could underestimate muscular challenges when strong co-activation is present. Nevertheless, only data from agonist muscles were reported. Therefore, in this study, we recorded both agonist and antagonist muscles around selected joint to confirm the co-contraction phenomenon. Also, the ability of using these two ratios to quantify muscular challenges in young and elderly adults was assessed

CLINICAL SIGNIFICANCE

Muscle strength of the lower extremities is essential for balance maintenance and mobility. Reduced muscular strength is one of the physical changes that can significantly impact an older adult's functional ability (Buchner, & de Lateur, 1990). Therefore, quantification of the muscular demand during gait is needed for better understanding of age-related decline in muscle function.

METHODS

Ten healthy elderly adults (HE) (mean age = 70.7 ± 3.3 years, 172.7 ± 8.6 cm, 76.8 ± 15.2 kg) and ten young healthy adults (HY) (mean age = 21.9 ± 2.0 years, 174.4 ± 10.1 cm, 71.1 ± 17.1 kg) were recruited in this study. EMG activities of the vastus lateralis (VL), semitendinosus (ST), medial head of gastrocnemius (GA), and tibialis anterior (TA) muscles of the dominant leg were monitored. Joint moments during MVC for the hip abductors, adductors; knee extensors, flexors; ankle plantar, and dorsi flexors were determined using the Kinetic Communicator Exercise System (KinCom) (Chattecx, Hixson, TN). Subjects were asked to walk on a level surface and stepping over the obstacles corresponding 2.5% and 10% of each individual's body height. Whole body motion was captured with an 8-camera motion analysis system (Motion Analysis Corp., Santa Rosa, CA). A total of 29 markers were placed on the subject's bony landmarks (Chou et al. 2003). Two force plates (AMTI, Watertown, MA) were used to collect ground reaction forces. Each gait cycle of the trailing and leading limb was divided into four phases. The N-EMG and N-moment ratio were calculated throughout the gait cycle.

Two-way ANOVA with repeated measure was used to detect the effects of age and obstacle height on all variables. The significance levels for all tests were set at 0.05.

RESULTS

HE demonstrated a trend of having greater N-moment ratios of all joints than HY in both leading and trailing limbs. However, only the hip abductor N-moment ratios reached statistically group differences ($p < .05$). Except the knee extensor moment ratio of the leading limb during early stance(Fig 1.), greater N-moment ratios were found for all joints in both trailing and leading limbs when crossing over a higher obstacle ($p < .05$).

Compare to HY, HE showed significantly greater N-EMG ratios in all 4 muscles of the trailing limb ($p < .05$). When changing the testing limb as leading limb, HE demonstrated significantly greater GM, VL, and ST N-EMG ratios than HY ($p < .05$). Increasing obstacle height resulted in significantly greater N-EMG ratios for all muscles in both the trailing and leading limbs ($p < .05$; Fig 2.).

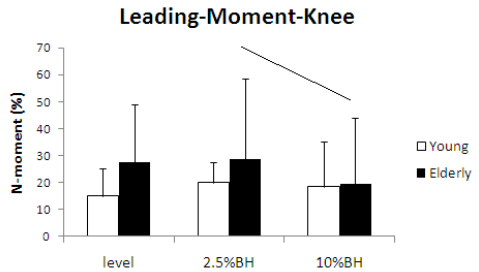

Fig 1. Peak knee extensor N-moment of the leading limb.

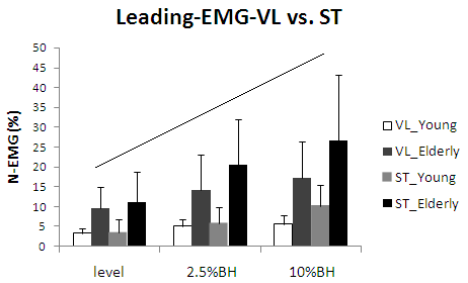

Fig 2. Mean vastus lateralis normalized EMG activation ratio during single stance phase (phase III) of the leading limb.

DISCUSSION/CONCLUSIONS

In agreement with those reported previously (Chen, 2006; Lee, 2005), the leading limb knee extensor N-moment ratio decreased with the increasing obstacle height. However, the N-EMG ratios of VL and ST increased linearly with the increase of obstacle height in this study. This finding suggests that the muscular demand on the knee muscles are not diminishing as indicated by the leading limb knee extensor moment. When a strong co-activation between agonist and antagonist muscles is present, N-EMG ratio provided a better quantification of muscular challenges.

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DYNAMIC STABILITY OF EMG PATTERNS IN YOUNG AND ELDERLY GAIT

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INTRODUCTION

Every year, falls occur in over one-third of adults over age 65, and these falls are the primary cause of accidental death in this population [1,2]. A person must generate both the correct motor patterns to facilitate stable walking, and also appropriate corrective responses to both large and small perturbations. To understand how the nervous system performs these functions, we need to know how motor patterns vary during walking. However, much is not known about how EMG patterns from one stride relate to the next stride, or how they interact with other muscles. We compared the muscle activation patterns between young and older adults by describing those muscle activation patterns in state-space to account for the multivariate dynamics of muscle activation patterns over multiple consecutive strides.

CLINICAL SIGNIFICANCE

Gait occurs over multiple strides, yet most traditional gait analysis methods do not consider the multivariate, time-series nature of EMG patterns during gait. By describing effects of healthy aging on the spatio-temporal dynamics of EMG patterns we can better understand the processes of disease and the onset of functional decline in older adults.

METHODS

Eighteen healthy older adults (age 72±6) and 17 height- and weight-matched young adults (age 23±3), with no orthopedic or neurological conditions, participated with informed consent. Each subject completed 2 trials of 5 minutes each walking on a Woodway treadmill at each of 5 speeds, ranging from 80% to 120% of their preferred walking speed (1.0×PWS) [4]. Surface electromyography (EMG) was recorded (Delsys Inc., Boston, MA) from the *v. lateralis* (VL), *b. femoris* (BF), *gastrocnemius* (GA), *t. anterior* (TA) of the left leg at 1080Hz. Linear envelopes were calculated after bandpass filtering (20-300 Hz), rectifying and smoothing using a convolution filter. Inter-stride variability (mean standard deviations), local dynamic stability (divergence exponents) and orbital stability (maximum Floquet multipliers) were calculated from the EMG linear envelopes [4]. Local and orbital stability analyses were based on defining an 8-dimensional muscle activation state space for each trial that consisted of the four normalized EMG values and their respective time derivatives. Time derivatives were included to also capture the *rates* of muscle activation dynamics. State spaces The two groups and 5 walking speeds were compared using ANOVA (SPSS 14) .

RESULTS

Preferred walking speeds (PWS) in these very healthy elderly (1.29±0.15 m/s) and young (1.30±0.10 m/s) subjects were not significantly different ($p = 0.86$) [4]. In all subjects, variability of EMG linear envelopes increased with walking speed in VL, BF, and TA ($p < 0.001$), and in GS ($p = 0.027$). However, despite walking at similar speeds, EMG variability was larger in older adults for BF ($p = 0.003$) and TA ($p < 0.001$) muscles, but not for VL ($p = 0.180$) or GS ($p = 0.407$). Older adults also demonstrated greater local (Fig. 1) and orbital *in*stability of their EMG patterns. Both short-term (λ^*_S) and long-term (λ^*_L) local divergence exponents (Fig. 1) increased with speed ($p < 0.001$), while λ^*_S was greater (i.e., more unstable) in older adults ($p < 0.003$). Maximum FM were also larger (i.e., less stable) in older adults than in young ($p < 0.009$; not shown). FM did not vary significantly with speed ($p > 0.15$).

SUMMARY/CONCLUSIONS

EMG patterns in older adults exhibited greater local and orbital instability, similar to kinematics [4]. Thus, EMG amplitude patterns were more sensitive to small inherent perturbations in older adults, *independent* of walking speed. Thus, muscle activation patterns may be more easily perturbed in older adults, which may explain the similar kinematics results [4]. Older adults exhibit increased EMG response latencies to larger perturbations [5]. The increased λ^*_S and maximum FM observed during *un*perturbed walking may reflect similarly slowed responses to even very *small* perturbations. Responses to small inherent perturbations observed in standing posture can be predicted using a pinned-polymer model [6]. A similar relationship may also exist for walking. This needs to be confirmed in future work.

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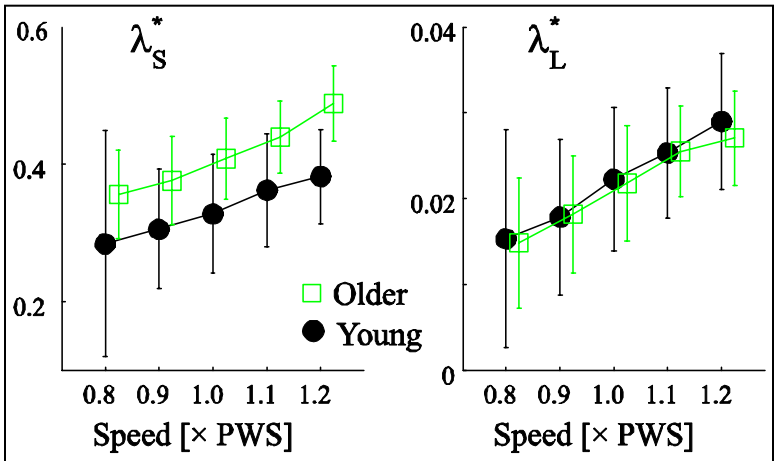

Fig. 1. Local Divergence Exponents vs. Gait Speed. Inter-subject means are shown. Error bars denote between-subject standard deviations within each group. Local dynamic instability increased with faster speeds, and older adults displayed larger λ^*_S than younger adults.

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The Influence of Heel Height on Utilized Coefficient of Friction during Walking

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Introduction

Wearing high heeled shoes has been associated with several clinical conditions (i.e. knee pain and low back pain), as well as increased potential for slips and falls.[1,2] Slips occur when the utilized friction (uCOF) of an individual exceeds the available friction provided by the shoe/ floor interface (available friction).[3] The association between wearing high heels and the increased potential for slipping suggests that the friction demand while wearing high heels may be higher when compared to wearing low heeled shoes. Although the kinematics and kinetics of high-heeled gait have been well described,[1,2,4] no study has examined the uCOF when ambulating in shoes of varying heel heights. The purpose of this study was to determine if heel height affects uCOF during walking. A secondary purpose of this study was to identify the factors contributing to altered uCOF when wearing shoes of varying height (i.e. walking velocity and ground reaction forces)

Clinical Significance

The results of our investigation may provide insight on whether wearing high heeled shoes increases an individual's risk of slipping. Data obtained from this study may highlight the need for the design of high heel shoes that are more slip resistant.

Methods

Fifteen healthy women (mean age 24.5 ± 2.5 yrs) participated in this study. Subjects walked at self-selected velocity under 3 different shoe conditions that varied in heel height (low 1.27 cm, medium 6.35 cm, and high 9.53 cm). Each subject was provided with footwear in their respective size. All 3 shoes had the same manufacturer and were chosen for their similarities in design, construction materials, and quality. Ground reaction forces were recorded using a force platform (Advanced Mechanical Technologies, Inc., Newton, MA) at 1560 Hz. uCOF was calculated as the ratio of resultant shear force to vertical force. For each trial, subjects' peak uCOF was determined during the first 50% of the stance phase. A one-way repeated measures ANOVA was performed to test for differences in peak uCOF among the 3 shoe conditions. Similarly one-way ANOVA's were performed to assess for differences in walking velocity, as well as the vertical force and resultant shear force at the time of peak uCOF. For all ANOVA tests, *post hoc* comparisons consisting of paired t-tests were employed using a Bonferroni adjustment. All significance levels were set at $p<0.05$.

Results

The ANOVA comparing peak uCOF among the 3 shoe conditions was significant ($p<0.001$). On average, significantly greater uCOF values were observed with increasing heel height (Fig. 1). Walking velocity decreased with increasing heel height, with statistically

significant differences being observed between each shoe condition. With respect to the ground reaction forces at the time of peak uCOF, significant reductions in the vertical forces were observed with increasing heel height (p=0.007). In contrast, there was a corresponding increase in the resultant shear force as heel height increased (p=0.007).

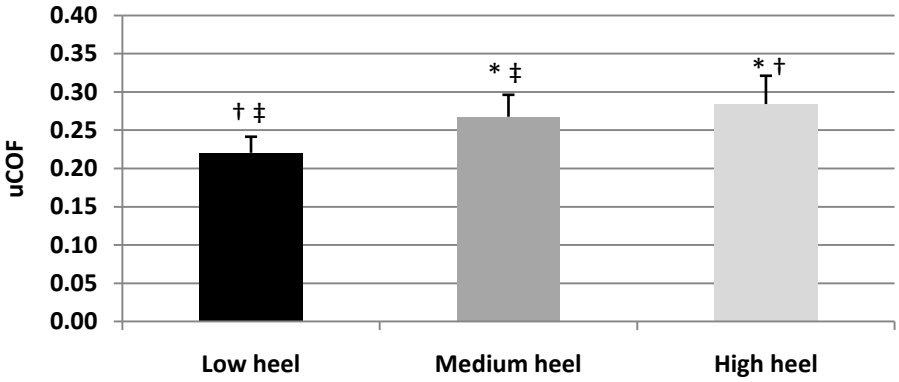

Fig 1. Comparison of peak uCOF across the 3 heel height conditions. All height comparisons were found to be statistically significant.

Discussion

Our results indicate that as shoe heel height increases, the friction demand during walking increases. This finding reinforces the premise that one is more likely to slip when walking in a high heeled shoe compared to when wearing a low heeled shoe. For example, it has been reported that when an individual's uCOF exceeds the available friction of the shoe/floor interface by as little as 0.006, slip risk can increase by 50%.[5] uCOF increased with heel height despite the fact that walking velocity decreased while wearing the higher heeled shoes. In general, the increased uCOF with increasing heel height was the result of an increase in the resultant shear force and a concurrent decrease in the vertical force. Our results signify the need for proper public education and increased footwear industry awareness of how high heeled shoes may affect slip risk.

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EFFECT OF SURFACE INCLINE ON MULTI-SEGMENT FOOT KINEMATICS

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INTRODUCTION

Ramps and surface inclines encountered during community ambulation may be functionally more challenging than level ground walking, particularly in subjects with foot pathology with limited motion. Lower extremity changes during incline walking have been previously reported by McIntosh et al in 2006. In 2002, Urry described foot pressure changes during side slope walking. However, the specific effects of surface incline on multi-segment foot kinematics remain unclear. Understanding the mechanism of adaptation to surface inclines in the healthy foot may provide a means to better identify and describe pathological foot motion in the future.

CLINICAL SIGNIFICANGE

There is a lack of understanding of the foot mechanisms involved in adapting to changes in surface incline. In the pathological foot, there may be reduced ability to alter foot motion to accommodate surface incline changes, which may lead additional gait disturbances during community ambulation.

METHODS AND PROCEDURES

Ten adults (average age 23.9 ± 3.7yrs) underwent gait analysis with IRB informed consent. Each subject was instrumented with a modified Helen Hayes marker set, and an eight marker multi-segment foot marker set (Tulchin, 2004). Marker trajectories were collected using a VICON motion capture system at 120Hz and filtered with a Woltring spline with MSE of 10. Subjects walked on a treadmill at 3.0mph at inclines of 0% to 12% grade in 3% increments. Decline trials were conducted with the treadmill at 3.0mph by placing wooden blocks under the rear of the treadmill, at approximately -7.5% grade. ANOVA's with Scheffe's test post hoc were used to detect differences between grades. It was hypothesized that there would be significant differences in hindfoot and forefoot sagittal motion with changes in treadmill grade.

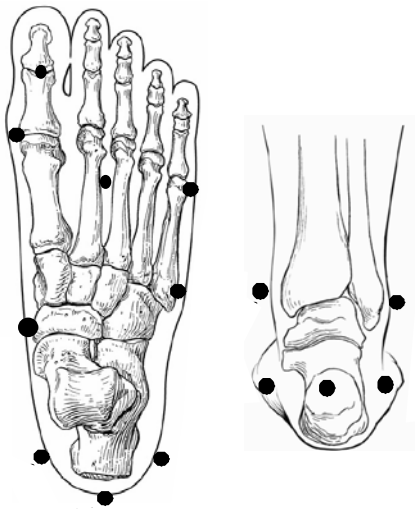

Figure 1: Skin marker placement for the foot kinematic model.

RESULTS

Significant differences were seen in sagittal plane hindfoot motion between inclines. Maximum hindfoot plantarflexion during first rocker was significantly less for the 9% and 12% grades as compared to decline walking, level walking and lower incline conditions ($p<0.0001$), Table 1. The timing of peak hindfoot dorsiflexion occurred significantly earlier in the gait cycle during the higher incline conditions. ($p<0.0001$) For healthy adults, there were no significant differences in sagittal plane forefoot motion or timing of peak motion.

	Decline	Level	Incline			
	~ -7.5%	0%	3%	6%	9%	12%
Max HF PF 1st rocker (°)	5.8 ± 4.0 [9, 12%]	4.3 ± 4.0 [9, 12%]	4.0 ± 3.9 [9, 12%]	4.0 ± 5.8 [9, 12%]	0.9 ± 4.3	-0.4 ± 3.4
Max HF DF 2nd rocker (°)	7.6 ± 3.0 [12%]	8.1 ± 3.2	8.7 ± 3.2	8.7 ± 3.0	9.4 ± 3.6	10.2 ± 3.5
Time Max HF DF 2nd rocker (%gc)	42 ± 6 [6, 9, 12%]	39 ± 3 [9, 12%]	40 ± 3 [9, 12%]	38 ± 4 [9, 12%]	34 ± 6	32 ± 6

Table 1: Significant differences were seen in hindfoot sagittal plane motion between incline conditions. mean ± stdev. ** Significant difference from the condition shown by superscript

DISCUSSION

Incline does appear to affect foot kinematics, particularly at the hindfoot. Specifically, the hindfoot has been shown to achieve less dorsiflexion during decline walking, and more dorsiflexion during incline walking during the first half of stance phase. The timing of peak sagittal hindfoot motion occurs earlier in the gait cycle as treadmill incline increases. These surface incline changes in foot segment kinematics have not been previously reported.

Interestingly, the sagittal plane forefoot motion did not change with varying grades of treadmill walking. This implies that the ankle is able to accommodate the changes in grade and the forefoot simply follows along.

This ability to alter ankle or hindfoot motion may not be possible in the pathological foot during community ambulation, where subjects are likely to encounter ramps and other surface inclines. This may lead to changes in forefoot motion not seen in the healthy foot, and warrants further investigation.

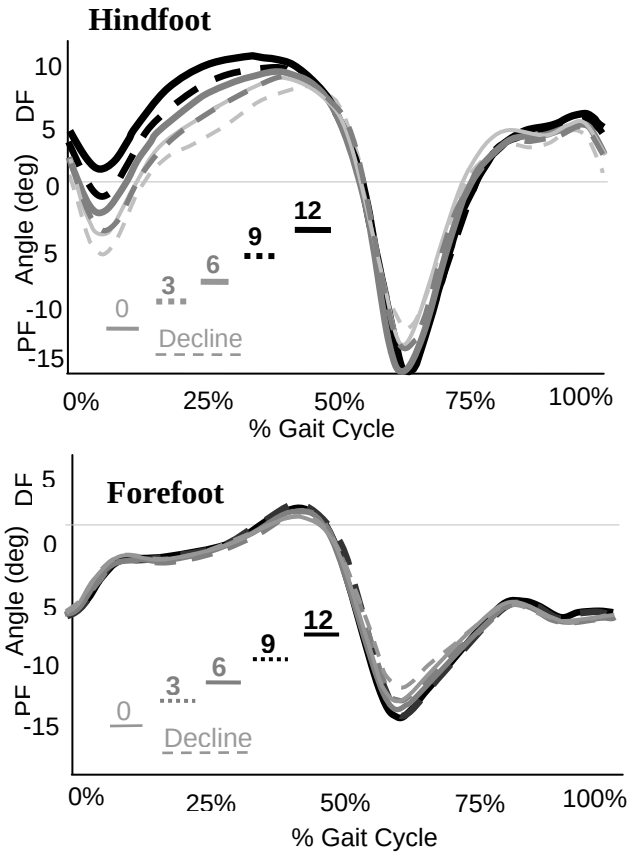

Figure 2 Significant changes in sagittal plane hindfoot motion were seen with incline treadmill walking, however there were no changes in sagittal plane forefoot motion.

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SUPPORT SURFACE AND VISUAL FIELD TRANSLATIONS AFFECT VARIABILITY AND DYNAMIC STABILITY OF WALKING DIFFERENTLY

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INTRODUCTION

Improving the ability to react to perturbations during daily activities like walking is key to effective fall prevention training. To assess training effectiveness, we must first quantify how walking stability changes in response to de-stabilizing events. In robotics, walking stability is commonly quantified using Floquet multipliers (FM) [1]. Healthy elderly exhibited larger maximum FM (i.e., more unstable) than younger subjects during *un*perturbed walking [2]. However, FM did *not* increase when perturbations of increasing amplitude were applied to a dynamic model of walking [3]. The purpose of this study was to determine how continuous horizontal pseudo-random oscillatory perturbations affect walking stability in *humans*.

CLINICAL SIGNIFICANCE

Virtual reality (VR) environments provide a unique opportunity to apply well-controlled and repeatable visual and/or mechanical perturbations during gait. Applying such perturbations through a VR system would enhance gait rehabilitation by creating realistic and challenging walking scenarios that can still be executed in the safety of the clinic.

METHODS

Twelve young healthy subjects participated with informed consent. Each subject completed five 3-min walking trials in a Computer Assisted Rehabilitation ENvironment (CAREN) under each of 5 conditions: no perturbations (NOP), anterior-posterior platform (APP) or visual (APV) translations, or mediolateral platform (MLP) or visual (MLV) translations. All translations were pseudo-random sums of sines with frequencies of 0.16, 0.21, 0.24 and 0.49 Hz. Kinematic data for the head, trunk, pelvis and feet were collected using Vicon at 60 Hz.

Means and standard deviations of step length (SL), step time (ST) and step width (SW) were computed for each trial. To quantify orbital stability, maximum FM were computed for each time series using established techniques [1,2,3]. State-spaces were generated using 3-dimensional velocities and accelerations of the C7 vertebral marker. The maximum FM defines the amount by which a perturbation away from the mean reference trajectory grows or decays after one complete cycle. If the maximum FM has magnitude < 1, perturbations decay after successive strides, and the system is orbitally stable. Smaller FM indicate greater stability. The results were analyzed using 2-factor (Condition $\times$ Subject) ANOVAs.

RESULTS

Mean SL and ST were significantly shorter during the ML perturbations than during either the AP or NOP conditions ($p < 0.001$). Mean SW and SW variability were both significantly

larger during the ML perturbations than during either the AP perturbations or NOP condition ($p < 0.001$; Fig. 1). Conversely, however, subjects remained orbitally *stable* (maximum FM < 1) for *all* experimental conditions. Maximum FM magnitudes did not increase significantly ($p \geq 0.403$; Fig. 1) for any of the perturbation conditions relative to the NOP condition.

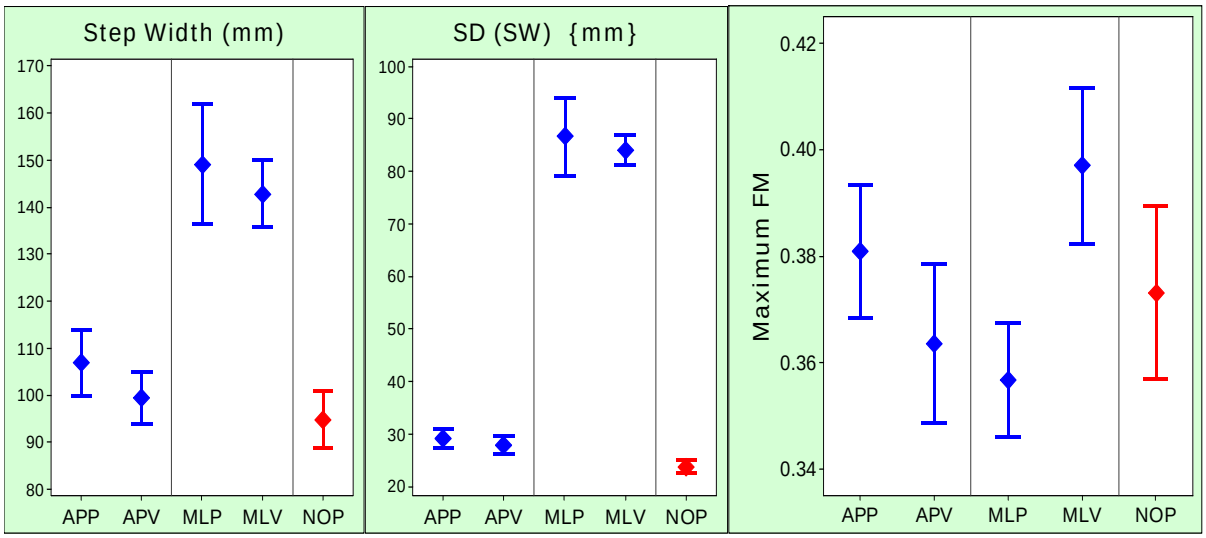

Fig. 1. Mean SW, standard deviation of SW, and maximum FM for each of the perturbation conditions (blue) vs. the NOP condition (red). Error bars indicate ± 95% confidence intervals.

DISCUSSION

Subjects were more sensitive to platform perturbations than to visual perturbations. Subjects were also far more sensitive to ML perturbations than to AP perturbations, consistent with theoretical predictions [4]. During the ML perturbations, subjects took shorter, faster, wider, and more variable steps. However, even with these substantial perturbations, subjects still maintained stable walking for all conditions (no subject stumbled or fell). Inducing greater movement, and thus greater movement *variability* does not necessarily induce greater *instability*. The FMs theoretically reflect the inherent stability of the *system*. This should not depend strongly on the details of the perturbations applied. This is reflected in the stable FM values (Fig. 1) and directly confirms modeling predictions, where orbital stability did not change as perturbation amplitudes increased [3]. More work is needed to determine how people *achieved* stability and decreased their risk of falling, in spite of increased variability.

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ACKNOWLEDGEMENTS

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DO HUMANS OPTIMALLY EXPLOIT REDUNDANCY TO CONTROL STEP VARIABILITY DURING WALKING?

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INTRODUCTION

Humans typically choose step length and frequency combinations that minimize energy cost [1]. However, such optimizations only predict *average* behavior and do *not* explain the considerable stride-to-stride *variability* observed in walking. The human nervous system either must *overcome* variability as a limiting constraint [2], or may *exploit* variability to optimize task performance [3]. Defining the essential *goal* of a task [3,4] can distinguish these alternatives. One primary goal of walking on a motorized treadmill is to maintain, *on average* and *over time* (i.e., *not* necessarily at each stride!), constant speed. We used a specific decomposition approach [6] to determine how humans adjust stride-to-stride movements during treadmill walking.

CLINICAL SIGNIFICANCE

Increased gait variability has been linked to increased fall risk in the elderly. Thus, it is important to understand precisely how humans *regulate* stride-to-stride variability in walking.

METHODS

Seventeen healthy young (age 18-28) volunteers walked on a level treadmill (Woodway USA) at their preferred speed. The 3D movements of reflective markers attached to their feet were recorded continuously using Vicon for 2 trials of 5 minutes and used to compute the stride length (L_n) and stride time (T_n) for each stride, n.

The *goal* of steady-state walking is to maintain constant speed, v. However, *many* combinations of stride length (L_n) & stride time (T_n) can achieve this goal. Specifically, all combinations of $L_n/T_n = v$ define a "Goal Equivalent Manifold" (GEM) [6] for walking (Fig. 1). We therefore decomposed variability in L_n and T_n into variability tangent to $(\hat{e}_T)$ and perpendicular to the GEM $(\hat{e}_P)$ via:

$$\vec{\delta}_n = \delta_T\,\hat{e}_T + \delta_T\,\hat{e}_P \qquad (1)$$

Scalar deviations δ_T do *not* affect walking speed. However, deviations δ_P directly represent *errors* in walking speed. We thus extracted time series of deviations tangent (δ_T) and perpendicular (δ_P) to each GEM for each trial (Equ. 1; Fig. 1).

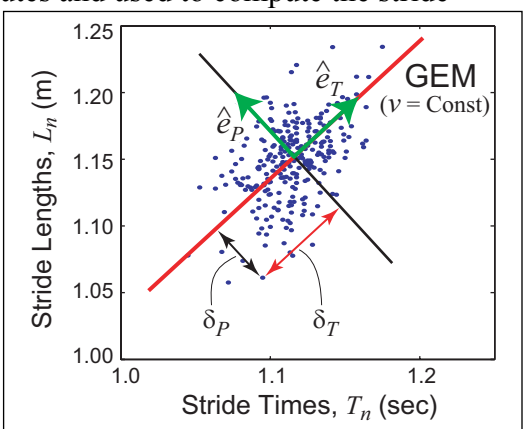

Fig. 1: Stride times (T_n) and lengths (L_n) for a typical subject, showing the GEM and scalar deviations δ_T and δ_P.

Detrended fluctuation analysis [5] was used to calculate stride-to-stride scaling exponents, α. A value of α > 0.5 indicates statistical "*persistence*" (i.e., deviations go *un*corrected over consecu-

tive strides). Conversely, $\alpha < 0.5$ indicates "*anti-persistence*" (i.e., deviations are corrected rapidly).

RESULTS

Subjects exhibited significantly less variability perpendicular to the GEM (Fig. 2A "OR"). They exhibited persistent correlations ($\alpha > 0.5$) for δ_T deviations *along* the GEM, but *anti*-persistence ($\alpha < 0.5$) for δ_P deviations *perpendicular to* the GEM (Fig. 2B "OR").

Randomizing the original L_n and T_n time series to create randomly shuffled (SH) or phase-randomized (PR) surrogates [6] yielded significantly *different* dynamics than original data (OR). Thus, variations in OR were *not* random and independent of the GEM.

We then implemented a strict "minimum intervention principle" (MIP) model [3]. This also yielded dynamics significantly different from the original data (OR). Defining a "preferred operating point" (POP) on the GEM for the model (analogous to optimizing energy cost) improved performance (Fig. 2), but still did not account for the *anti*-persistence seen in the δ_P deviations. Only a *sub*-optimal model (OVC) that slightly *over*-corrected for deviations away from the GEM exhibited *anti*-persistence in δ_P, similar to humans.

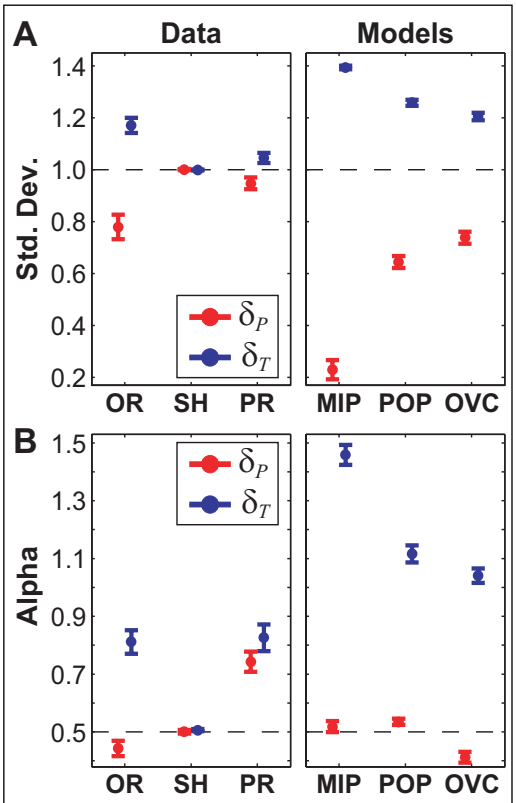

Fig. 2: **A)** Standard deviations and **B)** DFA Alpha (α) values for δ_P and δ_T deviations for the original (OR) and surrogate (SH, PR) data sets and also for the 3 models (MIP, POP, and OVC). Error bars = 95% confidence intervals.

SUMMARY/CONCLUSIONS

Humans explicitly adjust stride-to-stride movements to maintain constant speed at *each* stride, precisely as predicted [4]. Alternative (i.e., surrogate) strategies that would have equally accomplished the walking task, but with *no* explicit reference to the GEM, were *not* used. Finally, stochastic optimal control models demonstrated that healthy humans did *not* adopt precisely "optimal" [2,3] strategies, but instead exhibited *sub*-optimal performance where they consistently *over*-correct for small deviations in walking speed from each stride to the next.

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SHOULDER DEMANDS ACROSS MANUAL WHEELCHAIR ACTIVITIES

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INTRODUCTION

The shoulder joint is the most common site of upper extremity (UE) pain in manual wheelchair users, with the reported incidence ranging from 32%-78% [1]. UE intersegmental kinetics have been examined during steady-state ergometer and level propulsion to explore the connection between mechanical loads during wheelchair propulsion and UE pathology[2, 3]. However, no study has compared starting, stopping, and steady-state propulsion with high demand activities. The purpose of this study was to investigate shoulder joint kinetics over a range of daily activity and mobility tasks associated with manual wheelchair propulsion.

CLINICAL SIGNIFICANCE

It is critical to the understanding of UE pathology in manual wheelchair users to fully investigate the activities that may be associated with pain and injury. These data provide characterization and comparisons of previously unexamined conditions.

METHODS

Twelve subjects (mean age of 43 years, 11 males, 1 female) with at least 6 months experience using a manual wheelchair were recruited for this study. Eleven subjects were manual wheelchair users secondary to spinal cord injury (T4 lesion level and below); one secondary to spina bifida. Subjects used their own wheelchair for all testing, and both wheels were removed and replaced with SmartWheels (Three River Holdings, LLC, Mesa, AZ). Subjects performed four propulsion tasks and a weight-relief task. The propulsion tasks were over a level surface, starting from rest, stopping, and up a 10 m, 1:12 ramp. Subjects performed three trials for each task and were instructed to propel at a natural speed.

Kinematics were recorded with a 10-camera Motion Analysis EVa RealTime system (Motion Analysis Corp., Santa Clara, CA). Thirty-two retro reflective markers were placed bilaterally on the subjects' UE. The SmartWheels collected 3D handrim forces and moments about the hub of the wheel. Kinematic and kinetic data were recorded simultaneously at 240 Hz.

The kinematics and handrim kinetics were inputs into an upper extremity model built in Visual 3D (C-Motion, Inc., Rockville, MD). Inverse dynamics was used to determine the intersegmental joint forces and moments at the shoulder. Forces were expressed in the humerus coordinate system and normalized to body mass for analysis. Moments were expressed in the non-orthogonal joint coordinate system: plane of elevation (POE) flexion/extension, elevation adduction/abduction, and axial rotation [4]. Moments were normalized to body mass and height. Peak values were identified in each force and moment direction. A one-way ANOVA with five repeated measures (level, ramp, start, stop, and

weight relief) was performed for each peak kinetic value. When significant main effects were identified, post hoc Bonferroni-adjusted pair wise comparisons ($\alpha \leq .05$) were performed.

RESULTS

The anterior, posterior, medial, and superior force directions; and the POE extension (ext), elevation adduction (elev add), and external rotation (ext rot) moments resulted in significant effects across conditions (Table 1). The remaining force and moment directions were not significant across conditions. Weight relief resulted in significantly higher forces and ramp propulsion resulted in significantly higher moments than the other conditions. Surprisingly, the start condition resulted in large muscular demands about the shoulder equivalent with that of ramp propulsion, while the demand imparted by the stop condition was shown to be equivalent to level propulsion across all forces and moments.

Table 1. Peak Shoulder Joint Kinetics Across Conditions

Kinetics	Level	Ramp	WR	Start	Stop	P-value
Forces (N/kg)						
Anterior	0.23±0.21†	0.65±0.20‡	0.09±0.27†	0.23±0.36†	0.07±0.56†	0.001*
Posterior	0.72±0.22†	1.70±0.24‡	1.91±0.76‡	1.10±0.84†	0.57±0.45†	<.001*
Medial	0.15±0.25‡	0.49±0.57†,‡	0.96±0.73†	0.09±0.47‡	0.23±0.80‡	0.003*
Superior	1.21±0.35‡,§	1.89±0.58‡	4.10±1.28†	1.22±0.32‡,§	0.83±0.31§	<.001*
Moments (Nm/kg*m)						
POE ext	0.06±0.03†	0.13±0.05†,‡	0.22±0.13§	0.17±0.09‡,§	0.08±0.04†,‡	<.001*
Elev add	0.05±0.02‡	0.14±0.04†	0.06±0.04†,‡	0.13±0.12†,‡	0.07±0.08†,‡	0.009*
Ext rot	0.04±0.04†	0.15±0.10‡	0.11±0.07†,‡	0.17±0.08‡	0.08±0.12†,‡	0.004*

Values are mean ± SD; *Sig difference between conditions (P<.05); $\dagger$, $\ddagger$, §: statistical groupings

SUMMARY/CONCLUSIONS

This study describes shoulder demands across a spectrum of activities and provides data on conditions not previously characterized. Weight relief and propulsion initiation resulted in intersegmental kinetics that are potentially injurious. Further study is necessary to link activities to injury and to develop recommendations for propulsion and activity modifications that will mitigate harmful high joint forces and moments to reduce injury risk.

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GRIP FORCE PRODUCTION IN THE ASSISTING HAND ON PERFORMANCE OF A BIMANUAL TASK: APPLICATIONS TO HEMIPLEGIC CEREBRAL PALSY

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INTRODUCTION

Patterns of precision grip force production during unilateral grip/lift tasks have been used to elucidate typical sensorimotor mechanisms required for object manipulation and to identify grip force impairments in clinical populations, including hemiplegic cerebral palsy (HCP)[1]. The purposes of this ongoing pilot project are to 1) develop an alternate grip force paradigm that measures force production in the non-dominant hand as it assists in a bimanual task; 2) identify typical patterns of grip force output; and 3) test applicability to children with HCP.

CLINICAL SIGNIFICANCE

Current interventions for children with HCP emphasize improved assistive functions of the affected hand in bimanual tasks. Grip force measures describe interaction of the assisting hand with objects and may be a useful complement to upper extremity motion analysis.

METHODS

Twelve children with typical development between 5 and 12 years of age (9.09 ± 2.25) and seven with HCP between 5 and 12 years (8.38 ± 2.25), who demonstrated at least minimal voluntary grip and release and no significant joint limitations, participated in the first phase of the project. The experimental task required the subject's non-dominant hand to reach for and grip an instrumented "cup", transport it to the workspace of the dominant hand, hold it as the dominant hand placed tiny objects into the cup's well, place the cup, and release it. The shallow well design constrained subjects to orient the cup near vertical to place and maintain the objects successfully. Maximum isometric grip forces were tested separately using the cup.

The custom plastic "cup" is rectangular (5 x 8.9 x 2.5 cm), weighs 0.55 lbs and houses a 4 cm tri-axial force transducer (ATI Industrial Automation) which records normal forces exerted by the thumb as it opposes the fingers to grip. Aluminum plates (3 mm thick) covered with moleskin provide grip surfaces for the thumb and opposing fingers. A shallow well (0.9 cm deep) at the top of the cup holds tiny objects (1.1 cm x 0.5 cm). The grip apparatus also includes a signal conditioning box (ATI), laptop computer, data acquisition card (National Instruments) and Labview data acquisition software (National Instruments). The paradigm also involves three-dimensional kinematic analysis recorded with a Vicon MX motion capture system, but these data will not be addressed in this report. Simple kinematic trajectories were used to track the position but only to designate events in the force trace. Videos of the trials performed by children with HCP were reviewed by two investigators.

RESULTS

For all typical participants, standard events were identified in the grip force profiles of the non-dominant hand during performance of the bimanual task (Figure 1). Typical grip

force measures structured by these standard events are reported in Table 1.

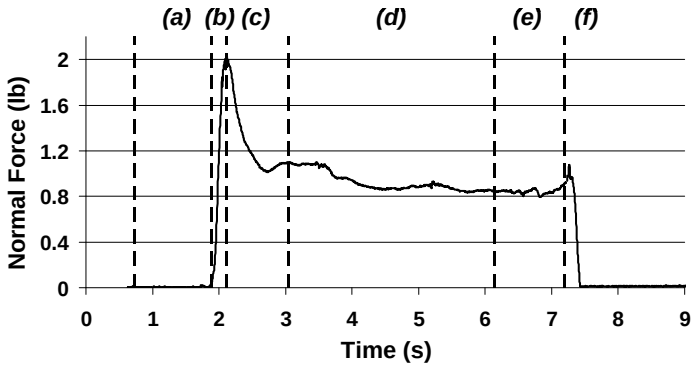

Figure 1. Events in a typical grip force trace in the non-dominant hand of a 12 year old participant: reach (a), grip (b), transport (c), hold (d), place (e), and release (f).

Table 1. Selected grip force measures averaged across all typical trials (n=12)

Grip Force Measure	Functional Event	Mean (SD)
Peak force (lbs)	Initial Grip	1.73 (0.79)
Rate of force generation (lbs/s)	Initial Grip	10.99 (7.58)
Mean grip force (lbs)	Holding	1.04 (0.35)
Coefficient of Variation	Holding	0.081 (0.040)
Rate of force decline (lbs/s)	Single Peak Release	-5.98 (3.98)

Seven children with HCP demonstrated individual differences in grip force events and measures that lead to inferences about force production deficits. These differences were interpreted in the context of proximal movement patterns identified on video review. Selected measures on five of the cases are reported below to exemplify the application of the paradigm.
Case 1: ↓ coordination of force generation. Trial #1: initial grip/holding not achieved. Trial #4: holding achieved with improved mean grip (1.55 lbs, or 33% of maximum isometric grip).
Case 2: Unsteady force magnitude/proximal instability. Trial #1: increased CV (0.154) and increased mean grip (3.23 lbs, or 50% of maximum isometric grip) for holding.
Case 3: Unsteady grip force/distal dystonia. Trial #1: increased CV (0.250) and mean force (1.94 lbs, or 36% of maximum isometric grip) in holding. Variability in subsequent trials.
Case 4: ↓ release of grip force/poor placement with proximal compensations. Trial #1: multiple peaks on place/release. Trial #5: Single peak release with -1.43 lbs/s decline rate.
Case 5: Relatively intact grip events/constraints in approach to object. Trial #1: relatively mild reduction in grip force generation (5.29 lbs/s) and increased CV (0.106) for holding.

DISCUSSION

Data from the first phase of this pilot study support further study of this novel force paradigm to advance understanding of grip force impairments in HCP. Future work will address kinematic analyses and quantification of aberrant force profiles.

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UPPER EXTREMITY KINETICS AND OUTCOMES ASSESSMENT DURING CRUTCH-ASSISTED GAIT IN CHILDREN WITH MYELOMENINGOCELE

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INTRODUCTION: Children with myelomeningocele (MM) make up a significant portion of the estimated 36,000 pediatric crutch users in the United States [1]. Often these children develop upper limb impairments that may lead to later conditions such as osteoarthritis, which may decrease their functional ability. Thus, we would like to study the relationship between upper extremity (UE) dynamics and functional outcomes. We hypothesize that UE dynamics during Lofstrand crutch-assisted gait are strongly correlated to the Pediatric Outcomes Data Collection Instrument (PODCI) standardized outcomes tool. The current study investigates UE joint dynamics and PODCI outcomes in 9 children with MM. Upper extremity dynamics of reciprocal and swing-through gait demonstrate important correlations with functional outcomes.

CLINICAL SIGNIFICANCE: Myelomeningocele (MM) is complex disease that often results in lower extremity weakness. It has been reported that approximately 50-60% of children with MM ambulate in the community. Twenty-three percent (23%) of these children use an assistive device, such as Lofstrand crutches [2]. During Lofstrand crutch-assisted gait a substantial amount of body weight is placed on the UEs. Literature has shown that longer-term crutch usage may result in UE pathologies, such as rapidly progressive destructive shoulder arthropathy, degenerative arthritis, or carpal tunnel syndrome [3,4].

METHODS: Nine subjects (11.1 ± 3.8 years) with L3 or L4 level myelodysplasia participated in the prospective study. Children were evaluated with the PODCI for 1) upper extremity and physical function, 2) transfers and basic mobility, 3) sports and physical function, 4) pain and comfort, and 5) global function and symptoms (Table 1). Motion analysis was conducted to evaluate UE dynamics using our model with seven rigid body segments and 26 markers. Crutches were instrumented with 6-axis dynamometers to measure applied reaction forces along the Cartesian X, Y, and Z-axes. The UE model was previously evaluated for accuracy and precision [5]. Subjects walked at a self-selected speed with each crutch pattern (reciprocal and swing-through) during multiple trials. Inverse dynamics was used to calculate UE kinetic measures.

Table 1. Mean (std) scores of the PODCI

PODCI Category	Mean Score
UE and Physical Function	93.7 (6.9)
Transfers and Basic Mobility	80.6 (7.9)
Sports and Physical Function	52.4 (17.9)
Pain and Comfort	90.2 (12.1)
Global Function and Symptoms	79.2 (6.4)

RESULTS: Overall, few significant correlations ($R > 0.7$ and $p < 0.05$) were seen amongst joint dynamics and functional outcomes (Table 2). Of these, most were unilateral. The highest number of correlations occurred between dynamics and upper extremity and physical function.

Table 1. Summary of correlations between dynamic metrics and PODCI (ROM: range of motion, FTI: force-time integral, MIF: maximum inferior force, MFM: maximum flexion moment, MEM: maximum extension moment)

PODCI Category	ROM	FTI	MIF	MFM	when MFM occurred	MEM	when MEM occurred
UE and Physical Function	S, E, W	E, W	W				
Transfers and Basic Mobility		S		E, W	W		S
Sports and Physical Function				S, E			
Pain and Comfort			W			S	S
Global Function and Symptoms		S, E, W		S, E			

Significant correlations ($R > 0.7$ and $p < 0.05$): S = shoulder, E = elbow, W = wrist

DISCUSSION: This study is the most extensive, to date, report of upper extremity dynamics during Lofstrand crutch-assisted gait. Currently, there are no reports in the literature that quantify the relationships among UE dynamics during Lofstrand crutch-assisted gait in children with MM and clinical outcomes measures.

While the current study showed few significant correlations between joint dynamics and functional outcomes, several key exceptions were noted. Crutch range of motion during reciprocal and swing-through gait was directly correlated to UE and physical function. The measured mechanical impulse during both gait patterns was strongly correlated to the outcomes measures of UE and physical function, and global function and symptoms.

In summary, this study identifies few significant correlations between joint dynamics and functional outcomes. Of those found, most were seen at the shoulder followed by the wrist and elbow. While the UE kinetic model did detect changes in joint dynamics, few changes were seen in the outcomes. Exploration of further, more sensitive outcomes measures may be suggested by the results of the current study.

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