O-01

KNEE EXTENSION AT TERMINAL SWING: A MISSING CRITICAL GAIT EVENT FOR CHILDREN WITH SPASTIC CEREBRAL PALSY

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Summary/conclusions
In a series of 175 children (276 legs) with spastic cerebral palsy (SCP) undergoing routine gait analysis, 87% of the legs demonstrated a limitation in peak knee extension at terminal swing (PKETSW) of 10° or greater. The degree of limitation in PKETSW was significantly different among three distinct patterns of initial foot contact, and significant relationships were detected between the knee extension limitation and kinematic variables linked to critical precursor events.

Introduction
Sufficient knee extension at terminal swing has been identified as a critical gait event because its existence positions the foot for weight acceptance [1]. Normal knee kinematics in swing reflect the behavior of the thigh and shank as a compound pendulum. Knee flexion in initial swing is promoted by factors that accelerate the thigh and flex the hip during limb advancement, but the sequence begins in terminal stance and preswing with ankle and hip power generation. Knee extension later in swing, is promoted by factors that decelerate the thigh at midswing and other complementary factors that grade the movement of the knee into extension [2, 3]. Symptoms of spasticity and impairments in selective motor control limit the precursor gait events that are required for normal PKETSW in children with SCP [3, 4]. The purposes of this retrospective study were to: 1) assess the prevalence of limitations in PKETSW, 2) examine the relationship of these limitations to pattern of foot contact; and 3) examine the association of kinematic variables that are linked to precursor events in the gait cycle to the limitations in PKETSW.

Statement of clinical significance
Limitations in PKETSW in children with SPC are highly prevalent, affect foot position at initial contact, and are linked to precursor critical events that may be limited by impairments in motor control. These findings may be useful in planning gait interventions for children with SPC and measuring the outcomes of the interventions.

Methods
Permission to conduct this retrospective study, under exempt status, was granted by the local institutional review board. Sagittal plane kinematic data from the initial gait analyses of all individuals with SPC, ages 5-21 years, conducted at the CGMA from June, 1999 to May, 2005 were entered into a database. About 58% of the samples were from children between 5 and 10 years of age; 27%, between 10 and 14 years; and 15%, between 16 and 21. The kinematic data were collected at 120 Hz using a 6-camera Vicon 512 motion capture system. Each leg constituted one sample. From the original 366 samples, 276 affected legs were maintained for analysis. Samples were omitted according to the following exclusion criteria: unaffected leg of a subject with spastic hemiplegia; use of assistive device during the gait trials; true equinus gait pattern, or crouch gait pattern. In true equinus and crouch gait, knee displacement are predictably constrained by the ankle or hip. Statistical analyses applied to the data are outlined in Table 1.

__Table 1.__ Guiding Questions and Statistical Analyses

Guiding Questions	Statistical Analysis
Prevalence of PKETSW	Frequency analysis; Descriptive Statistics
Effects of foot contact on degree of limitation in PKETSW	Analysis of Covariance (ANCOVA), with adjustment for peak ankle dorsiflexion in swing.
Relationship of kinematic variables to limitations in PKETSW	Univariate analysis (Pearson r correlation coefficients) to detect relationships of PKETSW with 17 relevant kinematic variables including peak values, % gait cycle and slope at hip, knee and ankle in swing period on ipsilateral side and in stance period on ipsilateral and contralateral sides. Multiple regression analysis of variables with Pearson r correlation coefficients > 0.30 (variables with fair-moderate associations), to assess their simultaneous effects on PKETSW

Results

Prevalence: Mean limitation in PKETSW, relative to normal (4° of flexion) was 21.35°±SD. Limitations in PKETSW of at least 10° relative to normal range were observed in 87.32 % of 276 samples. Of the 276 sampled, 12.68% demonstrated less than 10° of limitation; 35.14 %, a mild limitation (10°-20°); 33.70 %, a moderate limitation (20°-30°); and 18.48 %, a severe limitation (> 30°).

Effects of degree of limitation on pattern of foot contact: ANCOVA identified a significant effect of pattern of initial foot contact (forefoot, foot flat, or heel contact), with peak ankle dorsiflexion in swing accounted for ($p < 0.0001$). Post-hoc analysis identified significant differences ($p \leq 0.0065$) in PKETSW among all three foot contact patterns.

Relationships of kinematic variables: The multiple regression model accounted for 58% of the variance. Multiple regression analysis identified the following significant relationships: peak hip flexion in swing ($p < 0.0001$), slope of knee extension in swing ($p < 0.0001$), interaction of % gait cycle of peak knee flexion in swing with slope of knee extension ($p < 0.0001$) and with % gait cycle peak ankle dorsiflexion in stance ($p = 0.0006$).

Discussion

The kinematic variables and interactions that are significantly related to limitations in PKETSW, identified in the multivariate model, can be linked to three precursor gait events that affect knee behavior in swing; onset of plantarflexion in stance, control of forward thigh movement, and knee extension angular velocity at terminal swing. First, the onset of plantarflexion in stance is linked since timing of peak dorsiflexion interacts significantly with the timing of peak knee flexion in swing. Second, the control of forward thigh movement in midswing is implicated because of the significant relationship of peak hip flexion in swing and terminal knee extension limitations. Finally, the link to knee extension angular velocity at terminal swing is established by the significant relationship of the *slope* of knee extension and the excessive knee flexion just prior to initial contact, as well as the interaction of the slope with the timing of peak knee flexion in swing. Impairments associated with SCP may limit the motor control mechanisms required for performance of these precursor events at any stage prior to initial contact, which adversely affects foot placement. This analysis is the foundation for further investigation which will integrates kinematic with kinetic and clinical variables. The long-term objectives are to identify patterns among these variables and investigate causal relationships that will guide clinical decision making.

References

[1] Perry J, (1992), Gait Analysis: Normal and Pathological Function, 235-239.
[2] Davis RB & Õunpuu S, (2004), in Gage, The treatment of gait problems in cerebral palsy, 120-133.
[3] Carollo JJ, (2002), Phys Med & Rehabil Clin of N Amer, 13, 949-977.
[4] Cooney KM et al., (2006), Clin Biomech, 21, 59-66.

O-02

THE MODIFIED TARDIEU IN STANDING POSITION RESEMBLES GAIT CHARACTERISTICS MORE CLOSELY THAN THE SUPINE TARDIEU

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Summary

This study investigated the influence of the testing posture, knee velocity and hip fixation on spasticity assessment of the hamstrings using the Modified Tardieu Scale (MTS) in children with cerebral palsy and healthy controls. Moreover, the association between the MTS and the EMG-onset of the hamstrings during gait was tested.

Introduction

Gait abnormalities in children with cerebral palsy (CP), such as toe-walking and crouch gait, are often attributed to spasticity. However, the precise role of spasticity in disrupting gait has not been revealed. In order to quantify spasticity of the hamstrings, the Modified Tardieu Scale (MTS) has been widely used to compare the joint angle reached in a slow passive knee extension with the joint angle reached in a fast knee extension stretch. Several attempts were made to relate these measures to the EMG onset of the hamstrings during the swing phase of gait [1-2]. The success of these predictions was limited by difficulties in standardisation of the testing posture, controlling the hip angle and variations in velocity of knee extension[3].

In the present study, the MTS was performed with the patient in supine and (semi-supported) standing position. The standing position is hypothesized to be more similar to walking because of the presence of contralateral leg loading and a more physiological vestibulospinal drive. During assessment of the MTS, the consistency of the knee velocity and the movement of the hip have been investigated. Furthermore, the relationship between the EMG onset of the biceps femoris during the fast stretch and during gait was studied.

Statement of clinical significance

The MTS for the hamstrings in standing position with one leg supported resembles gait characteristics more closely than the MTS performed in supine position. Therefore, using the MTS in standing position might be advantageous compared to the supine testing posture for clinical assessment of spasticity.

Methods

In total, seven children diagnosed with CP (6.9 ± 1.6 years) and seven healthy controls (11.3 ± 2.4 years) participated in this study. In all children, the MTS was performed three times in supine position and three times in standing position with the investigated leg supported (hip 90° flexion; knee 90° flexion). Afterwards, the children walked at their preferred speed on a walkway. Outcomes of the most affected limbs of the CP group were compared with data from the left leg of the control group.

Sagittal kinematics of both knee and hip (goniometers) and EMG-data of the biceps femoris were collected during assessment of the MTS and walking. The angle subtended by the longitudinal axis of the tibia with the distally extended longitudinal axis of the femur was recorded as the popliteal angle of the knee joint. The movement of the hip during assessment was computed by subtracting the hip angle at the moment of the maximum knee angle from the hip angle before starting the stretch movement.

Results

The maximum knee velocities of the slow and fast stretch of the MTS were consistent and did not overlap. The maximum knee velocity of the fast stretch during standing position (417 ± 104°/s) closely matched maximum knee-extension velocity during gait (408 ± 122°/s), while the supine values (653 ± 90°/s) were approximately 50% higher on average. The changes of the hip angle during assessment of the fast stretch of the MTS in supine and standing position averaged 3.2 ± 6.5° and 2.2 ± 7.2°, respectively.

For the CP-group, the mean popliteal angle at which the biceps femoris was activated during gait was comparable to that assessed in the MTS in standing position (57.4 ± 9.1° and 59.2 ± 17.0°, respectively). In contrast, the mean popliteal angle observed in supine position (117.8 ± 9.5°) was much larger. For the control group the same tendency was shown, where the MTS in supine position (97.5 ± 13.8°) clearly overestimates the popliteal angle at biceps femoris onset in gait (50.6 ± 6.8°), while the angle at standing position (33.9 ± 14.7°) more closely matched with the gait values.

Discussion

Although the MTS in supine position is frequently applied in clinical practice, it may be better not to rely on the supine testing posture when using the MTS to predict hamstrings spasticity in gait. The present study showed that velocities of knee extension are considerably higher in supine position than those used during walking at preferred speed. In contrast, when the MTS is performed in a semi-supported condition (investigated side supported; contralateral side standing) the values for the knee velocity are much more similar to those at the end of swing phase in gait. The differences in maximum velocity may be related to more relaxation of the hamstrings in patients in supine position, to a difference in starting position of the knee angle and/or to the greater comfort for the experimenter to measure patients in the supine position.

Furthermore, the knee angle at EMG-onset of the biceps femoris in the MTS in standing position resembles gait characteristics more than the onset in supine position. Since gait deviations are proposed to be related to early or abnormal hamstrings activation in terminal swing-phase[1], the MTS in standing position seems preferable over the assessment in supine position. The similarity in EMG-onset at the MTS in standing position and gait for the CP group is likely to be due to the equivalent knee velocities and this finding underscores the importance of controlling knee velocity at the static tests.

Another important finding of the present study is that hip angles do indeed change during the execution of the test as was predicted[1]. Since changes in hip angle have a larger effect on the hamstrings muscle length than changes in the knee, it is quite important to take hip angle into account[4]. Hence, it is advised to fixate the hip and pelvis as much as possible during the test in order to minimize their contribution to the test-outcome.

References

[1] Damiano DL et al., *Gait Posture,* 2006, 23(1): 1-8
[2] Lebiedowska MK et al., *Arch Phys Med Rehabil*, 2004, 85: 875-80
[3] Scholtes VA et al., *Dev Med Child Neurol*, 2006, 48(1): 64-73
[4] Visser JJ et al,. *Eur J Appl Physiol Occup Physiol*, 1990, 61: 453-460

O-03

COORDINATION BETWEEN PELVIS, THORAX AND LEG MOVEMENTS IN GAIT

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Summary/conclusions

We analyzed the relationship between pelvis and thorax transverse rotations and the movements of the legs during gait in healthy volunteers. Transverse rotations of the thorax in healthy gait were out of phase with the leg movements at all velocities, whereas pelvis transverse rotations were in phase with the thorax rotation at low gait velocities and in phase with the leg movements at higher velocities.

Introduction

In normal gait, with increasing velocity the relative phase of pelvis and thorax rotations in the transverse plane increases, such that the coordination between these segments shifts from relatively in phase, towards a more out-of-phase pattern [1]. This phase shift is suppressed in several pathologies that affect locomotion, such as pelvic-girdle pain, low-back pain and Parkinson's disease [2,3,4]. Since many years, it has been assumed that in walking faster then 3 km/h pelvic step becomes important to launch the swinging leg, and that transverse counter rotation of the thorax is then needed to keep the angular momentum of the body around the vertical axis close to zero [5]. Therefore, the absence of a counter rotation of the thorax is thought to lead to inefficient gait. However, given the low inertia of the pelvis relative to the thorax, it seems unlikely that thorax rotation is aimed at compensating for pelvis rotation. We therefore, studied the coordination of thorax and pelvis rotations relative to leg movements.

Statement of clinical significance

Coordination between pelvis and thorax in gait is changed with disorders affecting locomotion. A better understanding of the nature and effects of these changes may refine diagnostics of gait disorders and give directions for interventions.

Methods

Nine healthy male volunteers walked on a treadmill at 9 different velocities, ranging from 2.0 km/h to 5.2 km/h, while 3D kinematics were recorded (Optotrak 3020, Northern Digital™, ON, Canada). From these data, relative timing between the pelvis and thorax rotations, between pelvis rotation and the thigh motion (sagittal plane motion of the distal right femur), and between thorax rotation and thigh motion was calculated as the Relative Fourier Phase (RFP) with 0 degrees being in phase and 180 degrees out of phase.

Results

The rotation of the thorax to the right (right shoulder backward) reached its maximum close to the instant of the maximum forward swing of the right thigh and vice versa for the rotation to the left, indicating a counter movement of the thorax relative to the legs at all velocities tested (Figure 1). The RFP between thorax rotation and thigh motion ranged from 157 (SD 18) degrees at the lowest velocity to 178 (SD 17) degrees at the highest velocity. The pelvis moved in synchrony with the thorax at low velocities and thus out of phase with the thigh (RFP: 112 (SD 18) degrees at 2.0 km/hr) and more or less in synchrony with the thigh, though still somewhat delayed (RFP: 52 (SD 33) degrees at 5.2 km/hr), at higher velocities (Figure 1). Consequently, the observed shift from in-phase towards out-of-phase pelvis-thorax coordination with increasing velocity (RFP: 44 (SD 18) degrees at 2.0 km/hr to 126 (SD 41) degrees at 5.2 km/hr) is mainly due to altered timing of pelvis rotations relative to the legs at higher velocities.

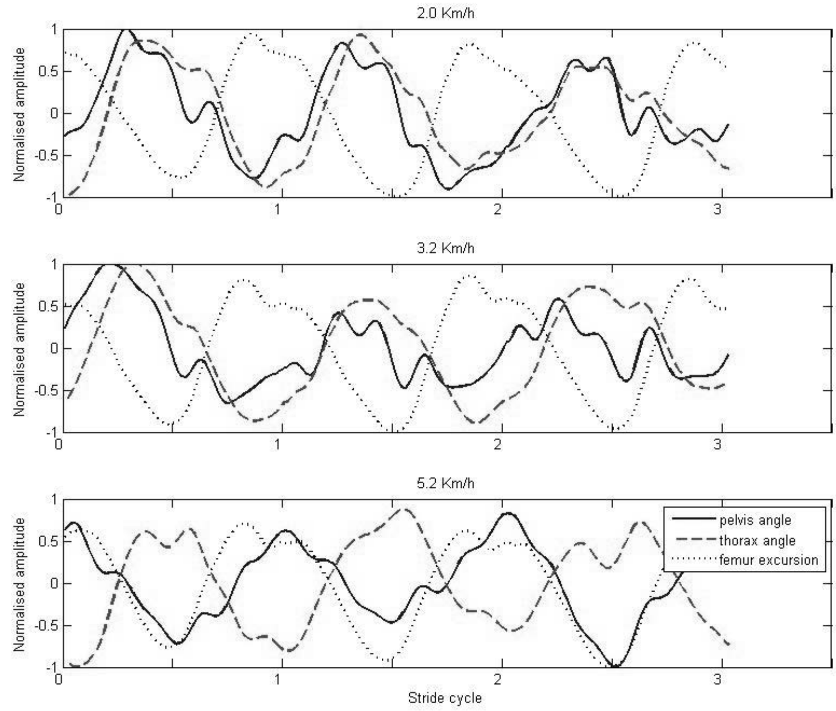

Figure 1. Typical example of the transverse rotations of thorax and pelvis and the forward/backward movement of the right distal femur in a healthy subject at three velocities

Discussion

The results presented suggest that thorax counter-rotation could be actively controlled to compensate the angular momentum around the vertical of the swing leg instead of the pelvis. Alternatively, it could be a consequence of the second-order dynamics of the system. At low velocity, thorax and pelvis may be strongly coupled. Counter-rotation with respect to the legs could then arise due to inertia of pelvis and thorax together. If, at higher velocities, the coupling between pelvis and thorax decreases, the system could be regarded as a system with two degrees of freedom (at the hips and the lumbar spine). This would lead to a decrease in pelvis-femur RFP, and an increase in pelvis thorax RFP. While the in-phase relationship between pelvis rotation and leg swing at high velocities contributes to step length, the pelvis rotation does not launch the leg since it follows the thigh movement at a slight delay. Preliminary analysis of data from patients with pelvic-girdle pain, low-back pain, and knee osteoarthritis indicate that changed pelvis thorax coordination can be caused by an alteration of the coordination of the pelvis relative to the leg movement, or by an alteration of thorax rotation relative to the pelvis. This suggests that different changes in gait coordination may lead to the same suppression of the phase shift of thorax relative to pelvis rotation with increasing gait velocity.

References

[1] Lamoth CJ, Beek PJ, Meijer OG (2002) Gait & Posture 16, 101-114.
[2] Lamoth CJ, Meijer OG, Wuisman PI, van Dieen JH, Levin MF, Beek PJ (2002). Spine 27, E92-99.
[3] Wu W, Meijer OG, Jutte PC, et al. (2002). Clinical Biomechanics 17**,** 678-686.
[4] Van Emmerik REA, Wagenaar RC, Winogrodzka A, Wolters EC (1999) Arch Phys Med Rehab 80, 186-191.
[5] DuCroquet R, DuCroquet J, DuCroquet P (1965). La Marche et les Boiteries. Paris: Masson.

O-04

CAN GAIT ANALYSIS GUIDE MANAGEMENT IN CHILDREN WITH SPASTIC DIPLEGIA?

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Summary
The short-term outcome in three groups of children with spastic diplegic cerebral palsy (SDCP), referred for three-dimensional gait analysis (3DGA) and treatment recommendations, was assessed retrospectively by looking at changes in the Normalcy Index[1] (NI) and minimum knee flexion in single support (MKFS) on a subsequent gait analysis. The groups consisted of 15 children who had multilevel surgery recommended and performed following 3DGA (operative group, OG), 15 children who had multilevel surgery recommended but not performed for family or administrative reasons (operation not done group, ONG), and 15 children in whom surgical intervention was not thought to be needed at that time (nonoperative group, NG) (Table 1). The NI results show that 3DGA was able to separate children with SDCP into distinct groups on the basis of severity of involvement. The NI in the OG decreased significantly between gait analyses, while the NI in the ONG and NG did not change significantly although subgroups showing improvement and deterioration were noted in both groups. The MKFS improved significantly in the OG between analyses, deteriorated significantly in the ONG, and did not change in the NG.

Introduction
Children with SDCP represent a heterogenous group, and 3DGA has the potential to identify the degree of involvement of a child with SDCP and thus inform clinical management. Concerns have been raised about the validity of treatment recommendations based on 3DGA in children with CP[2], and we wished to assess whether our recommendations were appropriate. Because of the ethical difficulties involved in performing a randomized controlled trial when surgical intervention may be indicated, we previously reviewed the outcome of our treatment decisions on the basis of the alteration in a number of kinematic variables between analyses. With the increasing use of the NI in assessing outcome[3], we repeated and expanded our study on expanded groups of children with SDCP consisting predominantly of children not previously assessed.

Statement of clinical significance
Three-dimensional gait analysis can effectively inform clinical management of children with SDCP by identifying children who would benefit from multi-level surgery and children in whom surgical intervention is not indicated. It can also demonstrate the heterogenous nature of SDCP and the variable natural history, which are important factors when considering the nature and timing of intervention and when interpreting the outcome of intervention.

Methods
The number of children available for the ONG and NG were limited, and we randomly selected an OG of similar size from the children who had attended the laboratory during the period of the study (Table 1). The gait analyses were performed by the same team using a Vicon 3D optical motion capture system. The treatment recommendations were made and the surgery performed by the same surgeon, and all children had similar postoperative rehabilitation. The children in the ONG and NG did not have any intervention apart from their usual physiotherapy during the study. The Normalcy Index (NI) was calculated retrospectively. The values for the reference and able-bodied groups used to form the NI were similar to published values[4]. We also looked at minimum knee flexion in single support (MKFS), which is not explicitly included in the NI.

Results
There were no significant differences in the ages of the children in the three groups. The initial NI values of the OG and ONG were similar, but the NI of both groups was significantly different from that of the NG ($p<0.001$). The same was noted with MKFS ($p<0.001$). The NI decreased significantly in the second analysis in the OG ($p<0.001$), but did not change significantly in the other groups.

Table 1: Details of study groups (see text for explanation of abbreviations)

	OG	***ONG***	***NG***
Mean age at referral (y), (SD)	8.88 (3.4)	10.13 (2.4)	8.73 (3.2)
Gender distribution (female, male)	9,6	7,8	2,13
Mean GMFCS level at referral	2.4	1.87	1.8
Mean interval between analyses (y) (SD)	1.69 (0.45)	1.16 (0.3)	1.28 (0.6)
Mean initial NI (SD)	1348 (708)	1079 (667)	260 (165)
Mean subsequent NI (SD)	746 (552)	1392 (1038)	239 (192)
Mean initial MKFS (SD)	26.5 (12.7)	23.9 (11.6)	5.5 (8.7)
Mean subsequent MKFS (SD)	12.2 (9.1)	34.3 (16.3)	5.1 (9.4)
Mean number of operative procedures recommended for each child (SD)	6.5 (2.4)	6.9 (2.2)	-

Within the ONG however, 9 children showed a deterioration in the NI between analyses, while the other 6 showed an improvement. Within the NG, 11 children had an improvement in their NI between analyses while the other 4 showed a deterioration. These subgroups of the ONG and NG did not differ in age, body mass index, GMFCS level, previous intervention, degree of fixed deformity, MKFS, or number of operative procedures recommended. The MKFS showed an improvement in the OG between analyses ($p=0.0001$), deteriorated in the ONG ($p=0.00001$) and remained similar in the NG ($p=0.81$).

Discussion
These results suggest that 3DGA can appropriately inform clinical management in children with SDCP in the short-term, and suggest that MKFS may have more of a predictive ability than the NI. The changes in the NI within the subgroups of the ONG are interesting and are not explained by the available data. Assuming that similar changes would have been noted in the OG if surgery had not been performed, it is possible that when assessing outcome we may be overestimating the effect of surgery in some children with SDCP and underestimating it in others. An increased use of 3DGA and the NI in children with SDCP may help us to define the natural history on an individual basis and thus optimize treatment recommendations and the timing of intervention.

References

1. An index for quantifying deviations from normal gait. Schutte LM, Narayanan U, Stout JL, Selber P, Gage JR, Schwartz MH. Gait Posture 2000;11:25-31.
2. Interobserver variability of gait analysis in patients with cerebral palsy. Noonan KJ et al, J Pediatr Orthop 2003; 23: 279-287.
3. Comprehensive treatment of ambulatory children with cerebral palsy: an outcome assessment. Schwartz MH, Viehweger E, Stout J, Novacheck TF, Gage JR. J Pediatr Orthop 2004:24: 45-53.
4. Use of the normalcy index for the evaluation of gait pathology. Romei M, Galli M, Motta F, Schwartz M, Crivellini M. Gait Posture 2004;19:85-90.

O-05

DISTAL FEMORAL EXTENTSION OSTEOTOMY AND PATELLAR TENDON ADVANCEMENT IN THE TREATMANT OF CROUCH GAIT: A REVIEW OF COMPLICATIONS

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Summary/Conclusions
The complication rate of patellar tendon advancement (PTA) is higher than that of distal femoral extension osteotomy (DFEO). The most common complications were different between the two procedures. Although complications can and do occur, with experience they can be anticipated, managed, and avoided.

Introduction
A clearer understanding of the dynamics of crouch gait is available today as a result of the study of typical and pathologic gait using modern motion analysis laboratories. Surgical advances in the techniques of distal femoral extension osteotomy and distal patellar tendon advancement have been previously demonstrated to be effective in the treatment of crouch and restoration of upright ambulation.[1] However the type and rate of complications of these procedures has not previously been examined. The purpose of this study was to examine the type and rate of complications associated with each procedure to assist in counseling families regarding the risks of intervention.

Statement of Clinical Significance
Because of their utility in managing crouch gait, DFEO and/or PTA procedures are becoming a common treatment recommendation for individuals with persistent crouch gait based on gait analysis findings. Until experience has been gained, new procedures can have a higher risk of complications. With practice, they can be anticipated, managed, and avoided. Report of complications and changes in techniques and care which have resulted are important education for others that may perform the procedures. Understanding the type and rate of complications associated with each procedure also enhances the ability to counsel patients and families regarding the risks of intervention.

Methods
A retrospective medical record review was done of all individuals who had undergone either DFEO, PTA or combination of the two procedures beginning with the first procedure which was performed in 1994 and the date chosen to end the study in June 2005. Data from the surgical admission to the last available outpatient clinic note was reviewed for each case. Any unwanted event was considered a complication. Short-term complications related to initial inpatient stay or casting were recorded but not considered as part of this analysis. Complication types and rates were recorded separately for each procedure in a consecutive manner.

Results
One hundred seventy-one individuals who had undergone one or both procedures were identified. A total of 205 surgical events were reviewed with 198 DFEO and 249 PTA performed. Sixty-seven patients were female, 104 were male. Their average age was 15.1±4.9 years (range: 4.6 to 39 years). Cases were performed by a total of eight surgeons. Seventy-eight percent of the cases were performed by two of the authors as first surgeon (50% jrg; 28% tfn). The majority of procedures were performed between 2000 – 2005 (88% DFEO; 90% PTA). The type of complications encountered for each procedure is found in Table 1.

Table 1: Type and Number of Complications

Complication Type	DFEO	PTA
wound dehiscence/infection	3	12
loss of fixation	1	10
persistent pain	3	9
nerve stretch/palsy/neuropathy	9	3
tibial tubercle fx	0	7
patellar tendon avulsion	0	7
post surgery deformity	5	1
non union/delayed union	2	4
arterial injury	1	0
miscellaneous	4	6
Total	**28**	**59**

Total complication rate for DFEO was 14% (28/198) and 24% for PTA (59/249). See Figures 1 and 2 respectively.

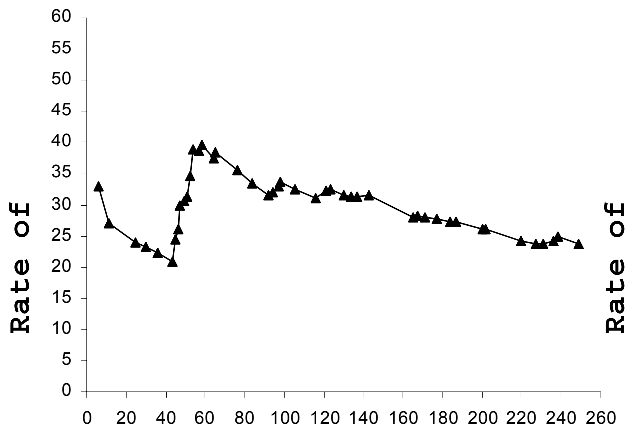

Figure 1: Patellar Tendon Advancement: Complication Rate vs. Procedures

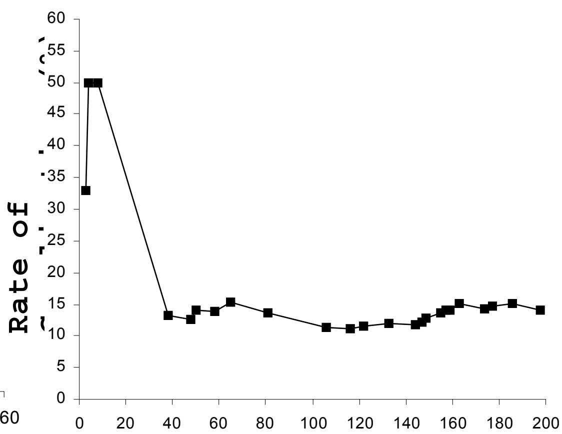

Figure 2: Distal Femoral Extension Osteotomy: Complication Rate vs. Procedures

Discussion

The most common complications were different between the two procedures. Complications associated with PTA were primarily related to wound infections and loss of fixation. Stretch palsy was the dominant complication with DFEO. Deformity recurrence occurred primarily when DFEO was performed in isolation early in the series. Many of the complications occurred in the first half of the series indicating a definite learning curve with these procedures. As a result, modifications of the surgical technique and post-operative management improved the safety and reliability of the procedures.

References

[1] Stout, JL, et al. Distal femoral extension osteotomy and patellar tendon advancement for treatment of persistent crouch gait in individuals with cerebral palsy, 2004, Proceedings of the European Society for Movement Analysis of Adults and Children, Warsaw, Poland

O-06

THE EFFECT OF SHOES ON GAIT IN CHILDREN WITH CEREBRAL PALSY

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Summary/conclusions

The effect of shoes on gait was evaluated for a group of children with spastic cerebral palsy (SCP) by comparing shod and barefoot walking. Walking with shoes significantly increased the stride length of children with CP, but did not significantly alter walking speed. Amongst the selected gait variables, shoes had a minor but significant effect on maximum hip extension and hip range of motion. Shoes had similar effects on gait to those reported in the literature for ankle foot orthoses (AFOs). Our results suggest that the effect of shoes on an individual patient's gait should be considered before AFOs are prescribed as a gait improvement measure.

Introduction

Children with CP demonstrate abnormal movement patterns secondary to a cerebral lesion acquired in their early development. Clinical gait analysis can be used to quantify abnormalities of the gait pattern. Although children in Western society normally wear shoes, common data collection in the laboratory is performed with the subjects walking barefoot. Any influence of shoes on their gait pattern is neglected in the analysis. Effects of AFOs have been reported on speed, stride length and cadence, whereas only minor changes have been shown on the kinematics [1-5]. But since AFOs can only be used in combination with shoes, the shoes may have contributed to the reported changes. The objective of this study was to assess the effect of shoes on spatio-temporal parameters, kinematics and muscle activity in children with cerebral palsy.

Statement of clinical significance

Shoes, worn without AFOs, contribute to an improved gait pattern in children with CP. Since the size of improvement is similar to those reported for AFOs, our results suggest that the role of AFOs as a gait improvement measure should be reconsidered.

Methods

Thirteen children with SCP (9 diplegia, 3 hemiplegia and 1 quadriplegia), aged between 7 and 16 years (mean 11.6 years) were recruited for this study. Walking was assessed in barefoot and in shoes in a randomised order using a 7-camera 3D motion analysis system. Surface EMG was recorded from *m. rectus femoris*, medial hamstrings, *m. gastrocnemius medialis* and *m. tibialis anterior* using surface EMG electrodes. In the trials with shoes, children wore their own (unmodified) shoes. N shod trials and N barefoot trials were selected for analysis. Cadence, speed stride length and step length were calculated from each trial, as well as maximum and minimum flexion angles of the knee and hip. The coactivation index between antagonist muscles was calculated according to Winter [6]. We used paired *t*-tests to ascertain if there were differences between the shod and barefoot groups (NS $p>0.05$).

Results

Significant increases in stride length ($p=0.0004$) and step length ($p<0.0001$) of 9 and 5 cm respectively were found in walking with shoes in children with CP. Only two significant changes in the kinematics were found: a decrease in minimum hip flexion during stance of 2º ($p=0.0001$) and an increase in the range of motion of 3º at the hip ($p=0.002$). The mean EMG amplitude and the level of coactivation over a gait cycle were not significantly different between the two conditions.

Discussion

The main result of this study was a mean increase in stride length of 9 cm in shod walking. This increase in stride length is in agreement with other studies that investigated the effects of shoes in stroke patients [7] and in typically-developing children [8]. The increase in stride length in the current study is probably due to the extra stability offered by the shoes. Stride length could be limited in children with CP because of poor balance perception or function. An increased stability could explain the preference of the children for walking in shoes (in the current study 8 out of 12 children preferred walking in shoes). Stride length is related to cadence and speed, but no significant changes were seen for these variables. It is possible that the power of the study was reduced by the limited number of children recruited into the study. No significant effects on EMG amplitude or coactivation were found for the musclses of the thigh or shank, suggesting that footwear does not moderate the effects of spasticity.

References

1. Abel *et al*. 1998. *Archives of physical medicine and rehabilitation* 79: 126-133.
2. Buckon *et al*. 2004. *Developmental Medicine & Child Neurology* 46: 590-598.
3. Dursun *et al*. 2002. *Disability and Rehabilitation* 24: 345-347.
4. Radtka *et al*. 2005. *Gait & Posture*; 21:303-10.
5. White *et al*. 2002. *Developmental Medicine & Child Neurology* 44: 227-232.
6. Winter, D.A. (1990). *Biomechanics and motor control of human movement*. New York: John Wiley & Sons, Inc.
7. Churchill *et al.*, 2003. *Clinical Rehabilitation* 17: 553-557.
8. Oeffinger et al., 1999. *Gait & Posture* 9: 95-100.

O-07

CORRELATIONS BETWEEN CLINICAL EXAMINATION AND GAIT DATA: IMPLICATIONS FOR CONTRACTURE CORRECTION

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Summary/conclusions
Contractures are treated to improve passive range of joint motion in order to optimise function. This study aims to investigate if there is a relationship between passive range and function. 14 subjects with cerebral palsy (CP) were recruited, all having passive range of knee extension increased with Contracture Correction Devices (CCDs). Passive range-of-motion data and other parameters from clinical examination were compared with gait kinematics and tempero-spatial parameters. Dynamic knee extension (at initial contact and mid-stance) and step length had a positive relationship with passive knee extension and a negative relationship with popliteal angle, as hypothesised. However, the correlation coefficients for the group were weak. The strength of the quadriceps had stronger correlations with gait parameters, suggesting that, to gain functional benefit from treating joint contractures, muscle strength must be maximised as well as passive range.

Introduction
Fixed joint contractures are a limitation of joint motion due to structural shortening. They can be treated by manual physiotherapy stretches, surgery, serial casting, and orthotics [1]. The latter can include the use of dynamic splints – orthoses with a spring mounted across the orthotic joint that provides a turning moment to the splint and hence a stretch to the contracted tissues [2]. The initial aim of all these treatments is to gain range of motion, and their success is usually judged by how much range increases. However, the ultimate aim should be an improvement in a patient's function.
The development of contractures occurs in response to immobility, which may be secondary to muscle imbalance, immobilisation, scaring, habitual postures and pain. The muscle imbalance, immobilisation & habitual postures can be due to spasticity, which is a common component of CP (cerebral palsy). When CP patients develop knee flexion contractures, these can have a significant effect on their functional mobility.
A study using gait analysis to monitor the effect of treating knee contractures with dynamic orthoses was undertaken which provided data on the relationship between clinical exam data (including passive range of motion) and function. Treating a knee flexion contracture is aimed primarily at increasing the range of motion of knee extension and extensibility of the hamstrings, but in ambulant patients it should also improve function. Therefore, it is hypothesised that, for children with spastic cerebral palsy who have knee flexion contractures, there is a correlation between passive range and functional kinematics. Specifically for treating knee flexion contractures, it is hypothesised that the knee flexion contracture angle has a negative correlation with knee extension at initial contact & mid-stance; that popliteal angle has a positive correlation with knee flexion at initial contact, but a negative correlation hip flexion at initial contact, and step length; and that passive ankle dorsiflexion has a positive correlation with dorsiflexion at initial contact and mid-stance.

Statement of clinical significance
The study tests if, for children with spastic cerebral palsy who have knee flexion contractures, there is a correlation between passive range of motion and functional gait parameters. If a strong relationship exists it would support the treatment of crouch gait by treating passive range.

Methods

14 subjects, all males mean age 14.2 years (SD 2.1) with a diagnosis of cerebral palsy spastic diplegia. Mean flexion contractures: hip=11.3º (7.3); knee=16.1º (8.9); ankle=3º (6.4). Mean popliteal angle=55.8 (6.9). Mean walking speed = 0.67m/s (0.25). All had treatment with a CCD for 1 hour daily (monitored by a data-logger) over a period of 12 weeks. Passive range & gait data were collected before and after treatment, and a record was also kept of additional factors such as growth, muscle powers and pain. Passive range of motion data was collected by the same clinician each time using a standard protocol and using the same design of hand held goniometer. Kinematic data was collected using a Kistler force plate (Switzerland) and a 6 camera VICON 612 motion analysis system (Oxford Metrics, UK).
Measurements of passive range of motion, growth, step length, and kinematic parameters were tabulated and the hypothesised relationships were tested by calculating correlation coefficients. Secondary parameters were analysed to identify additional important factors.

Results

Table 1. Correlation coefficients between Clinical exam data and Gait parameters

CLINICAL EXAM DATA	GAIT PARAMETERS					
	Hip flexion	Knee flexion		Ankle dorsiflexion		Step length
	IC	IC	MS	IC	MS	
Hip flexion contracture	0.031	-0.187	-0.067	-0.010	0.062	0.050
Popliteal angle	-0.164	0.249	0.324	-0.051	0.031	-0.317
Knee extension	0.331	-0.245	-0.286	0.139	0.076	0.379
Dorsiflexion knee extended	-0.102	-0.079	-0.037	0.163	0.249	0.152
Quads muscle grade	-0.258	-0.430	-0.537	-0.064	-0.172	0.481
Quads lag	0.009	0.422	0.367	0.098	0.096	-0.598

Discussion

Results show agreement with the hypothesised correlations, but for the group as a whole, the correlations are not strong. Additional factors were also tested for correlation (pain on walking, growth, and muscle strength). Muscle strength was measured by MRC muscle grading and lag (deficit in range from passive extension).These were found to correlate most strongly with gait data for the study group as a whole – specifically with knee flexion at mid stance and with step length – although correlation coefficients were no larger than -0.598. However, for individuals within the group there is strong agreement between PROM & kinematic measurements across consecutive visits, for example correlation coefficients between passive knee extension and active knee extension of greater than 0.9.
The correlation between quads strength and both functional knee extension and step length suggests that purely treating passive range of motion will not give as much functional benefit unless the treatment programme that also maximises muscle strength.

References

[1] Contractures in orthopaedic and neurological conditions: a review of causes and treatment. Farmer S E, James M, (2001), Disability and Rehabilitation, 23(13), 549-558

[2] Dynamic orthoses in the management of joint contracture Farmer SE, Woollam PJ, Patrick JH, Roberts AP, Bromwich W., (2005), Journal of Bone and Joint Surgery [Br], 87-B, 291-295

O-08

DOUBLE CALIBRATION VS GLOBAL OTPIMISATION: PERFORMANCE AND EFFECTIVENESS FOR CLINICAL APPLICATION

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Summary/conclusions

The performance of double calibration and global optimisation in reducing soft tissue artefact propagation to relevant joint kinematics was compared. For clinical application the quantification of actual subject-specific kinematics is necessary. Thus, according to what assessed, double calibration should be preferred.

Introduction

Soft Tissue Artefact (STA) was recognized the most critical source of error in clinical motion analysis [1]. A recent quantification [2] study has assessed marker displacements up to several centimetres resulting in large errors on knee rotations and translations, during the execution of several motor tasks, comparing stereophotogrammetric motion analysis results with a 3D fluoroscopic gold standard. In particular, STA propagation to knee kinematics resulted to make the quantified ab/adduction (AA) and internal/external rotation (IE) knee angles useless for clinical decision. Given the criticality of STA, several compensation methods were proposed in the literature [3]. Among these, recently proposed double calibration (DC) [4] was assessed to be very effective in the reduction of STA, allowing to calculate even reliable knee translations, comparable with the 3D fluoroscopic gold standard. On the other hand, the STA compensation method implemented in some commercial stereophotogrammetric systems is the global optimisation (GO) [5]. This method operates minimising the motion of the markers with respect to the underlying bony segments, assuming a predefined kinematic constraint at the joints. For clinical application, the STA compensation method applied should allow to quantify the actual kinematics of analysed subject. The aim of the present work was to assess the performance of DC and GO in quantifying subject-specific kinematics.

Statement of clinical significance

STA propagation to joint kinematics can nullify the clinical interpretability of stereophotogrammetric analysis [2]. STA was assessed to be strongly subject- and task-specific [2]. The present work provides relevant indication for the choice of the STA compensation method which allows to better reconstruct the specific kinematics.

Methods

The kinematic data-set was obtained by the synchronous combination of traditional stereophotogrammetry and 3D fluoroscopy analysis [2]. Data were obtained during the extension against gravity (EG), step-up/step-down (SUD), and sit-to-stand/stand-to-sit (STS) motor task from two subjects (P#1 and P#2) (age 67 and 64 years, height 155 and 164 cm, weight 58 and 60 Kg, Body Mass Index 24 and 22 kg/m2, follow-up 18 and 25 months) treated by total knee replacement. DC [4] and GO [5] methods were applied to stereophotogrammetric data, for the reconstruction of knee kinematics. The DC was performed interpolating two calibration configurations acquired at the extremes of the motion, for each motor task, with respect to knee flexion angle. The GO was applied assuming a ball and socket model for both the knee and the hip joint (according to [5] and most commercial applications). The knee kinematics reconstructed from 3D fluoroscopy was assumed as the reference gold standard [6]. The root mean square error (RMSE) of knee rotations and translations reconstructed using DC and OG was calculated over the repetitions for each subject and motor task with respect to the fluoroscopic gold-standard.

Results

Errors on knee rotations and translations reconstructed with OG were much larger than those obtained with DC.

Summarizing results are sketchy depicted in the bar plots reported in figure 1.

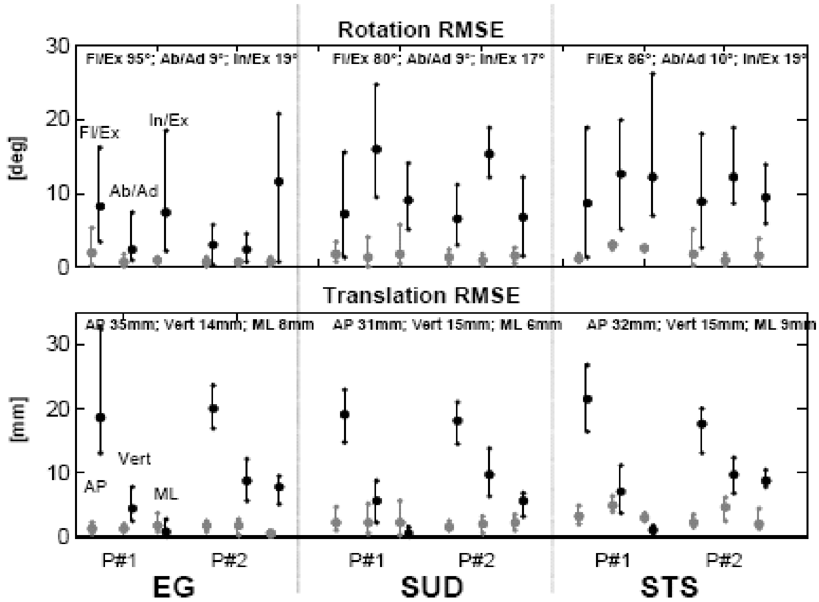

Figure 1. Bar representation of mean, maximum and minimum values over the repetitions of the RMSE on knee rotations (FE, AA, IE) and translations (AP, Vert, ML) reported for the two subjects (P#1, P#2) and three motor tasks (EG, STS, SUD). Results obtained from the application of DC (gray) and GO (black) calibration are reported. The mean reference range is reported on the top of the corresponding error bar plot for each motor task.

Discussion

The curves for knee rotations and translations obtained using GO are smooth and do not generally show non-physiologically wide ranges (e.g. AA and IE without compensation [2]). Nevertheless, as assessed by the large RMSE, these curves reproduce a kinematics which reflects the constraints imposed by the model, and not the kinematics of the specific subject analysed. On the other hand, DC, exploiting the subject-specific calibrations at the extremes of the motion, results to actually reduce STA, allowing to reconstruct the specific kinematics, without any pre-assumption. In conclusion, GO produces excellent joint kinematics for animation purpose, without the need of additional calibration, but for clinical application, where the subject-specific kinematics is necessary for evaluation purpose, DC should be chosen.

References

[1] Andriacchi, T. P. and Alexander, E. J., (2000), J.Biomech.,33, 1217-1224.
[2] Stagni, R. et al., (2005), Clin.Biomech.,20, 320-329.
[3] Leardini, A. et al., (2005), Gait Posture,21, 212-225.
[4] Cappello, A. et al., (2005), IEEE Trans.Biomed.Eng,52, 992-998.
[5] Lu, T. W. and O'Connor, J. J., (1999), J.Biomech.,32, 129-134.
[6] Zuffi, S. et al., (1999), IEEE Trans.Med.Imaging,18, 981-991.

O-09

THE SYMMETRICAL AXIS OF ROTATION APPROACH (SARA) FOR DETERMINATION OF JOINT AXES IN CLINICAL GAIT ANALYSIS

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Summary/conclusions
The goal of the study was to develop and evaluate a new technique, the Symmetrical Axis of Rotation approach (SARA) for determining the axis of rotation in e.g. knee joints during gait. This approach, capable of simultaneously considering two dynamic body segments, produced the best axis, when compared against literature methods using generated data. Application of this method in gait analysis should lead to improved clinical diagnosis and monitoring.

Introduction
Axes of rotation e.g. at the knee, are often generated from clinical gait analysis data to be used in the assessment of the severity of kinematic abnormalities, the diagnosis of disease, or the ongoing monitoring of a patient's condition. Currently available methods to describe joint motion from segment marker positions share the problem that when one segment is transformed into the coordinate system another, measurement errors and artefacts associated with motion of the markers relative to the bone are magnified [1]. We hypothesize that by calculating the axis of rotation using a method that can consider the movement of two dynamic body segments simultaneously, it should be possible to determine a unique axis of rotation more accurately than currently available methods. The goal of this study was therefore to develop a new robust approach that avoids the aforementioned problems and to compare its performance with a number of previously proposed techniques.

Statement of clinical significance
Gait analysis can allow the assessment and diagnosis of kinematic irregularities or deformities that may be unobservable even to a skilled clinician. Improvement in the accuracy of methods to determine the axis of rotation will enhance the assessment and monitoring of kinematic abnormalities.

Methods
A virtual hinge joint was created for which marker positions were generated computationally.
The two segments of the joint rotated around a common axis within a defined angular range of motion (RoM), resulting in circular plane arcs. On each segment, 4 markers were attached approximately 10 to 15 cm distant from the axis. Marker positions were then randomly distributed around the arc and within different RoMs. In addition, the whole joint configuration was able to randomly translate in space, simulating a non-stationary AoR. The following approaches were employed to determine the AoR:
- Algebraic Axis Fit (Circle fitting) [2]
    This technique presupposes that the AoR is fixed with respect to global coordinates.
- Mean Helical Axis [3]
    This algorithm assumes that one segment remains stationary and describes the movements as rotations around and translations parallel to the helical axis.

Transformation Methods: Transformation methods assume that the AoR is stationary only in each segment's local coordinates. Assuming at least 3 markers are present, then it is possible to transform from the global into time dependent local coordinate systems.
- Schwartz Transformation Technique (STT) [4]
    This approach determines the median of multiple pair-wise calculated axes.

- Symmetrical AoR Approach (SARA) (New)

The new SARA technique simultaneously considers that the axis parameters must remain constant in the local coordinate systems of both segments. Mathematically, this yields a linear least squares problem that has to be solved using singular value decomposition techniques to find the one-dimensional subspace corresponding to the smallest singular value which represents the joint axis.

To study the influence of skin elasticity and marker artefacts, different artificial error types were applied, based on in vivo data [1]. Isotropic, independent, and identically distributed Gaussian noise (SD 0.1 cm) was applied to each of the marker positions to represent elastic distortion of the skin. To model synchronous shifting of the skin over the underlying bones, the noise was also applied to the marker set collectively. All simulations were repeated 1000 times, each with 200 time frames.

Results

In general, the RMS error, a method of describing the quality of the calculated AoR, decreased for each approach with increasing range of movement. The SARA and Schwartz approaches together produced the smallest errors for all ranges of motion, but when additional simulated noise conditions were applied to each marker, the SARA alone determined the best AoR for two dynamic bodies.

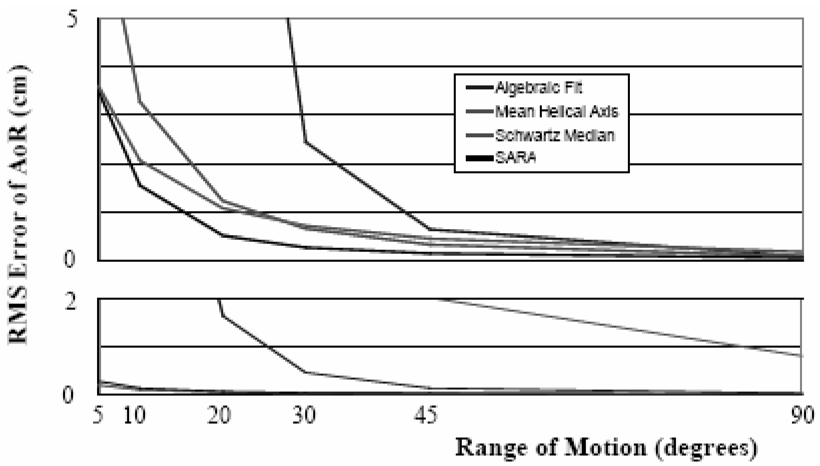

Figure 1. *RMS error (cm) of estimated AoR for different approaches over 1000 simulations. Top: Noise was applied to each marker of the segment set independently. Bottom: Noise was applied to the marker set collectively. A combination of these noise conditions produced approximately additive error magnitudes*

Discussion

The newly presented approach for determining the AoR, SARA, is a natural extension of the SCoRE algorithm [5] for determining joint centres, and is capable of considering two dynamic body segments simultaneously. Whilst initial results using the SARA are promising using numerically generated datasets and are fast enough to be determined 'on-line', the technique must now be proven in a clinical environment.

References

[1] Taylor Ehrig, Duda and Heller (2005), Journal of Orthopaedic Research, 23:4, pp 726-734
[2] Gamage and Lasenby, (2002), Journal of Biomechanics , 35:1, 87-93
[3] Woltring, (1990), in book "Data processing and error analysis", Berlec Corporation, pp 203-237
[4] Schwartz and Rozumalski, (2005), Journal of Biomechanics, 38:1, pp 107-116
[5] Ehrig, Taylor, Duda and Heller (2006) Journal of Biomechanics, in press

O-10

FEASIBILITY OF A NEW -JOINT CONSTRAINED- LOWER LIMB MODEL FOR GAIT ANALYSIS APPLICATION

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Summary/conclusions

A lower limb model has been developed in which kinematic constraints of hip, knee, and ankle joints are defined on the basis of functional anatomy and data collected from MRI and Fluoroscopy. In particular the femoral-tibial motion was supposed to be constrained by the knee cruciate ligaments. The feasibility of the model was checked on a normal subject walking on level at natural cadence and performing on site exercises. Thirty-one reflective markers were positioned on pelvis and lower limbs: they were used to identify anatomical landmarks and allowed us to connect a 3-D model of bones to the collected data. Our results show that kinematics of femur in relation to pelvis and shank can be accurately obtained through identification of hip joint centre, shank location, and knee joint kinematic constraints. The markers located on the thigh (greater trochanter, medial and lateral femoral epicondyles), which were the most affected by skin motion artefacts, can be profitably removed from our protocol, and the advantage will be reduced encumbrance and improved accuracy.

Introduction

Protocols for clinical gait analysis can be subdivided into those who don't impose any joint constraint (each segment is an independent free body in space), and those who pre-define a linkage between the anatomical segments. The advantage of the last ones is evident, in that the congruency of the relative movement of adjacent segments is inherently guaranteed, measurement errors of markers coordinates can be better distributed along the total limb, the problem of magnification of orientation errors of the local reference axes, because the markers are not positioned at the extremity of the anatomical segments, can be considerably reduced [1]. On the other side the joint-constrained models adopt very simple spherical or cylindrical hinges to describe the joint kinematics, which is inadequate particularly as far as the knee joint is considered. Due to this inadequacy the rigid body hypothesis cannot be satisfied, and considerable errors can be done in muscle-length estimation if muscle-tendon action lines are just attached to the skeleton model. Our new model attempts at overcoming the previous drawbacks by integrating imaging information into a model of the lower limb.

Statement of clinical significance

An improved identification of lower limb kinematics can be achieved trough proper modelling and anthropometric data collection from biomedical imaging. According to our scheme the markers on the thigh, which are the most affected by skin motion artefacts, can be avoided, and this can be an advantage in terms of reduced encumbrance and preparation time, that could be of interest for clinical application of gait analysis.

Methods

A motion analyser (SMART, eMotion, Italy) equipped with 6 TV-cameras located in a gait analysis laboratory was used for our experimental sessions. The model proposed was composed of four anatomical segments: pelvis, thigh, shank and foot. At variance with a previous protocol [2] the number of markers (31 in total) was redundant, in order to check for the relative inaccuracy. Five markers were located on the pelvis, and then, bilaterally, four on the thigh, five on the shank, and three on the foot. The hip, knee and ankle joint centres were tentatively identified as described in [3]. Then a 3-D model shank bones, obtained from previous elaboration of MRI [4] was adapted to the present subject by making them to best match anatomical landmarks. By an integrated analysis of MRI and Fluoroscopy (during knee flexion-extension), a four-bar linkage, in which the cruciate ligaments changed in length according to experimental data [4], was implemented and adapted to the present subject. The femur position was thus identified in each sampling frame as being defined by the position of the hip joint centre and by the cruciate ligaments. A best match procedure was implemented to this purpose. The analysed movements were: walking on level at natural speed, external/internal, abduction/adduction of hip with straight knees, crouching on the knees, standing up and sitting down.

Results

The kinematic variables obtained by both methods followed similar trajectories, nevertheless some divergences were found over the whole angular range of motion. At the knee, a good agreement among the flexion angle curves was found near maximum knee flexion, however a divergence was found that increased as the extension angle increased. Depending on the subject analysed, differences of up to 12 degrees in knee joint flexion/extension angle were obtained.

After defining the geometrical parameters of the model and attaching the movement analysis data, the initially estimated position of the hip joint centre was moved by the optimisation procedure used to match the model, within a range of 5mm.

Skin motion artefacts for the femoral condyle markers were found to reach 22.2mm for the full range of motion. Since the proposed approach does not use markers at condyles, this source of error was eliminated.

Discussion

By comparing the joint angles computed from the model and those obtained using the marker on the lateral femoral condyle the highest angular differences, of about 10°, were found at the stance phase, when full extension of the knee was observed.

After being applied in normal subjects, the method showed characteristics that make it adapted for advanced applications, e.g. applications related to skeletal-muscle system characterisation, starting from common movement analysis data without increasing the subject preparation complexity.

Since the coordinates of the attachment points of the ligaments on femur and tibia are the most affecting model parameters, anthropometrical variability as well as some other critical points, such as loading conditions, still need to be addressed.

References

[1] Lu T-W, O'Connor JJ (1999) Bone position estimation from skin marker co-ordinates using global optimisation with joint constraints. J Biomechanics 32:129-134

[2] Frigo C, Rabuffetti M, Kerrigan DC, Deming LC, Pedotti A. (1998) Functionally oriented and clinically feasible quantitative gait analysis method. Med Biol Eng Comput, 36(2), 179-185

[3] Frigo C, Rabuffetti M. (1998) Multifactorial estimation of hip and knee joint centres for clinical application of gait analysis. Gait and Posture, 8: 91-102

[4] Pavan EE, Pascolini G, Zappata A, Frigo C. (2005) Determination of knee joint kinematics by means of bio-images and modelling for clinical application of gait analysis. Siamoc 2005 Congress Abstracts

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O-11

A SINGLE GAIT CYCLE AS MEASURED BY FOUR CURRENT PROTOCOLS

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Summary/conclusions
A single gait cycle from a volunteer was analysed simultaneously by using four different protocols among the most commonly utilised in clinical gait analysis laboratories. The marker-set was designed to integrate those of the four relevant instructions. One volunteer was analysed, with marker placement and necessary measurements performed by relevant experts. Except biases associated to the different definitions of anatomical reference systems, the consistency among the four data set obtained was considerable.

Introduction
The large number of gait analysis laboratories all over the world routinely utilise data collection and reduction procedures embraced in a few well known protocols [1-5]. These differ considerably for the marker-set utilised, additional measurements taken at data collection, definition of the anatomical landmarks and references, joint axis conventions etc. Single cases or general studies are discussed with their clinical implications and shared at meetings and in journals indifferently coming from these disparate protocols without full consciousness of these methodological differences. In the present study, a comparison is made between four of these protocols by looking at exactly the same single gait cycle.

Statement of clinical significance
Gait analysis results are interpreted routinely within the process of clinical decision making, but it is still unknown to what extent the different protocols utilised worldwide compare to each other. This knowledge would make gait analyser more conscious about the reliability and the limits of his own clinical gait variables.

Methods
The Plug-in-gait (Vicon Motion Systems Ltd, Oxford, UK) [1,2], Saflo [3], CAST [4] and T3Dg [5] protocols were analysed. A single comprehensive marker-set was defined by integrating all the four marker-sets. This resulted in 59 markers: 22 in each leg, 5 in pelvis and 10 in trunk. One subject (male, 24 years, 188 cm, 74 Kg) was instrumented accordingly and was asked to walk barefoot at his normal speed. Necessary anthropometric measurements were performed by examiners skilled with these protocols. Marker trajectories and ground reaction force were collected respectively by an eight-camera Vicon 612 system and by two force plates (Kistler Instrument AG, Switzerland) at 100 Hz. A single representative trial including the right and left gait cycle was identified. Marker labelling and data processing including filtering were performed independently by the relevant experts. No normalisation or off-set removal was allowed. Relevant results were gathered from the four processes, with a careful comprehension of the correspondences. The variability between the protocols was determined for each kinematic and kinetic variable across the right gait cycle. The coefficient of correlation between variables obtained by the four protocols and the mean range of variability (MRV) were calculated, the latter being obtained by averaging the range of values at each sample along the gait cycle [6].

Results

MRV was found very small for pelvic rotation (1.1°) and obliquity (2.0°), and relatively small for knee Fl/Ex (7.9°). It was considerably large for hip Fl/Ex (20.2°) and ankle Do/Pl (18.3°) and Ab/Ad (15.8°) but this was expected because of the different definitions of the anatomical-based reference systems. Coefficients of correlation are reported in Table 1.

Table 1. Coefficients of correlation between the variables from couples of protocols.

Segment/Joint Rotations	T3Dg vs PiG	T3Dg vs Saflo	T3Dg vs CAST	PIG vs Saflo	PiG vs CAST	Saflo vs CAST
Pelvic Tilt	0,977	0,852	0,978	0,900	0,999	0,900
Pelvic Obliquity	0,991	0,670	0,973	0,727	0,992	0,804
Pelvic Rotation	0,997	0,986	0,996	0,727	0,992	0,993
Hip Flexion/Extension	0,994	0,998	0,999	0,994	0,995	0,998
Hip Abduction/Adduction	0,994	0,919	0,985	0,945	0,994	0,962
Hip Internal/External Rotation	0,269	-0,236	-0,382	-0,281	-0,164	0,882
Knee Flexion/Extension	0,992	0,984	0,995	0,994	0,995	0,995
Knee Abduction/Adduction	0,923	-0,169	-0,390	-0,356	-0,542	0,779
Knee Internal/External Rotation	-0,219	0,781	0,739	-0,090	-0,098	0,696
Ankle Dorsal/Plantar	0,949	0,960	0,974	0,986	0,976	0,979
Ankle Eversion/Inversion		0,024	0,950			-0,053
Ankle Adduction/Abduction		0,844	0,943			0,833
Joint Moments						
Hip Flexion/Extension	0,877	0,843	0,996	0,890	0,895	0,870
Hip Abduction/Adduction	0,970	0,988	0,997	0,972	0,968	0,989
Hip Internal/External Rotation	0,975		0,988		0,956	
Knee Flexion/Extension	0,958	0,937	0,976	0,963	0,970	0,987
Knee Abduction/Adduction	0,925	0,967	0,889	0,953	0,970	0,947
Knee Internal/External Rotation	0,957		0,969		0,975	
Ankle Dorsal/Plantar	0,999	0,999	0,999	0,999	0,997	0,998
Ankle Eversion/Inversion	0,529		0,991		0,583	
Ankle Adduction/Abduction			0,932			

Discussion

The MRV associated to the different protocols is not much larger than that associated to inter-observer and even inter-laboratory analyses [6,7] for most of the measurements. Correlation coefficients were found higher for kinetic than kinematic variables, and in the sagittal than out-of-sagittal planes.

References

[1] Kadaba MP, Ramakrishnan HK, Wootten ME. (1989) Measurement of lower extremity kinematics during level walking. J Orthop Res, 8, 383-392

[2] Davis RB III, Õunpuu S, Tyburski D, Gage JR. (1991) A gait data collection and reduction technique. Hum Mov Sci, 10, 575-587

[3] Frigo C, Rabuffetti M, Kerrigan DC, Deming LC, Pedotti A. (1998) Functionally oriented and clinically feasible quantitative gait analysis method. Med Biol Eng Comput, 36(2), 179-185

[4] Cappozzo A, Catani F, Della Croce U, Leardini A. (1995) Position and orientation in space of bones during movement: anatomical frame definition and determination. Clin Biomech, 10(4), 171-178

[5] Benedetti M.G., Sawacha Z., Paolini G., Giannini S., Leardini A. (2004) A new protocol for gait analysis in children. Gait & Posture 20(Suppl 1), S77

[6] Noonan KJ, Halliday S, Browne R, O'Brien S, Kayes K, Feinberg J. (2003) Interobserver variability of gait analysis in patients with cerebral palsy. J Pediatr Orthop, 23(3), 279-87

[7] Gorton G, Hebert D, Goode B. (2002) Assessment of the kinematic variability between twelve Shriners Motion Analysis Laboratories: part 2 short term follow up. Gait & Posture, 16(Suppl.1), S65-S66

O-12

IMPROVED TRACKING OF HIP ROTATION USING A PATELLA MARKER

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Summary/conclusions
A marker placed on the patella allows for more accurate measurement of hip rotations than traditional thigh wands. In controlled trials of isolated hip internal-external rotation, the patella marker detected 97% of the actual range of motion, compared with 59% for a distal thigh wand and 41% for a proximal thigh wand. The patella marker produced the smoothest hip rotation curves and the smallest hip rotation range in walking, and results from the patella marker did not depend on walking speed. These results suggest that the patella marker is less vulnerable to wobbling, inertial effects, and soft tissue movement than traditional thigh wands.

Introduction
Gait analysis is an important tool for treatment decision-making in orthopaedics. Hip rotation measurements from gait analysis are used in planning the femoral derotation osteotomy procedures. Accuracy in the measurement, therefore, is critical. Although thigh wand markers have traditionally been used to track hip motions, it has been reported that they produce large variability in hip rotation measurements between different laboratories and greatly underestimate the rotation movement mostly due to the existence of soft tissue artifacts [1,2]. We hypothesized that a marker placed over the patella would be less affected by soft tissue movement and would therefore produce more accurate hip rotation measurements. This study investigated effectiveness of the patella marker in comparison with traditional thigh wands.

Statement of clinical significance
If the patella marker provides more accurate hip rotation data than thigh wands, it will enhance the clinical usefulness of gait analysis, particularly in decision-making and surgical planning for femoral osteotomy.

Methods
Eleven healthy adults (Ht: 165.4 ± 7.6 cm, Wt: 63.2 ± 11.0 kg) performed controlled hip internal and external rotations while standing with 90-degrees of knee flexion and full hip extension to allow for calculation of a "gold standard" hip rotation [2]. Data were captured and quantified by a Vicon Motion Analysis System (Vicon Oxford Metrics, Limited, Oxford, England) using a standard Vicon marker set with knee alignment devices. The gold standard measurement (Gold) was calculated using additional markers placed on the fibular head and lateral malleolus to track rotation of the tibia. Three different thigh markers were used: a distal wand placed on the lateral aspect of thigh at 80% of total femur length (Dist), a proximal wand placed on the lateral aspect of thigh at 50% of total femur length (Prox), and a patella marker placed over the patella (Pat). Data were collected with all markers in place, and hip rotations for the same trial were calculated using the Gold, Dist, Prox, and Pat markers. Similar data were collected during walking, except that measurement of a gold standard was not possible.
The mean right hip rotation range of motion from each marker set was computed and compared to each other for both motions. One-way Repeated Measures ANOVAs and Scheffe's post hoc tests were utilized to determine the statistical significance.

Results

For the controlled hip rotation trials, the mean hip rotation ranges for Gold, Pat, Dist, and Prox were 58.7± 10.3°, 56.7 ± 9.8°, 34.2 ± 8.9°, and 23.7 ± 5.3°, respectively (Figure 1). Expressed as a percentage of the Gold Standard, the hip rotation ranges were 97%, 59%, and 41% for Pat, Dist, and Prox, respectively. The Statistics revealed significant differences ($P<0.0001$) between all conditions except Gold and Pat.

For walking, the mean hip rotation range was 13.5 ± 3.9° for Pat, 23.1 ± 4.7° for Dist, and 14.7 ± 8.0° for Prox (Figure 2). The results yielded a statistically higher mean ($P=0.0003$) in Dist compared to others. The hip rotation curves tended to be smoother for Pat compared with Dist and Prox (Figure 3). Hip rotation range tended to increase with walking speed for Dist, to a lesser extent for Prox, and not at all for Pat (Figure 4).

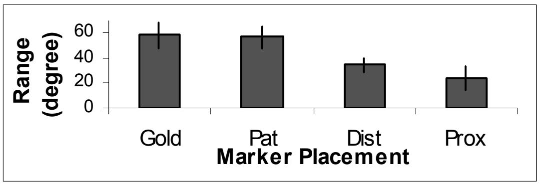

Fig 1. Hip rotation range during controlled trials

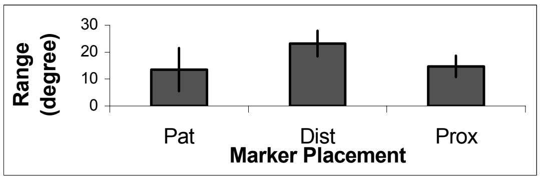

Fig 2. Mean hip rotation range during 1 gait cycle

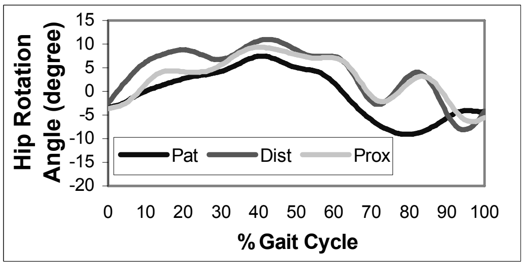

Fig 3. Hip rotation curves based on the three markers

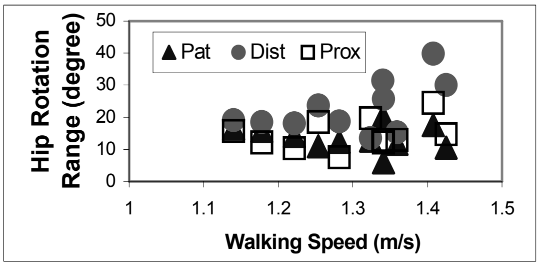

Fig 4. Hip rotation range as a function of velocity

Discussion

The patella marker demonstrated a greater potential in detecting hip rotation movement (97% of gold standard range) compared to the distal and proximal thigh wands (59% and 41% of gold standard range, respectively). The results for the thigh wands are similar to those reported previously by Lamoreux (70% for distal wand and 40% for proximal wand) [2]. Less soft tissue lying between the patella marker and the bone might have allowed the marker to track more direct movement of bone while the thick layer of soft tissues hindered this ability in the wands. It should be noted that the patella marker can only be used in conjunction with knee alignment devices.

In walking, the thigh wands produced relatively higher ranges of motion regardless of their poor rotation tracking capacity in the controlled tests. The rotation ranges of the wands, especially the distal wand, displayed sensitivity to walking speed. Their augmented ranges in walking, therefore, could possibly be explained by artifacts that are magnified with increased walking speed. The patella marker, on the other hand, was insensitive to walking speed and also produced the smoothest hip rotation curves. These results suggest that the patella marker is less vulnerable to wobbling, inertial effects, and soft tissue movement than traditional thigh wands.

References

[1] Gorton et al. (2001) Gait & Posture, 16 (S1), 65-66.
[2] Lamoreux (1991) Proceedings of the International Society of Biomechanics, Perth, 372-373.

Acknowledgements

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O-13

PROTOCOL CHANGES CAN IMPROVE THE RELIABILITY OF OXYGEN COST DATA

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Summary and Conclusions

Changes in data collection and processing methods substantially reduce the variability of net nondimensional oxygen cost (O_2 cost) among subjects with cerebral palsy (CP).

Introduction

The use of O_2 cost as a measure of gait efficiency is well accepted. There are concerns that the amount of variability in O_2 cost – both natural and due to experimental error –reduces its clinical utility [1]. Recently, a new scheme for normalizing O_2 cost data (net-nondimensional normalization) has been developed [2]. The net-nondimensional scheme has been shown to be superior to mass normalization in terms of removing covariate effects. One feature of the net-nondimensional scheme is the use of net (walking – resting) oxygen cost. The percentage variability in resting energy is significantly greater than in walking energy [3]. Furthermore, the experimental error in a difference of measures is inherently greater than for a single measure. Though natural variability cannot be controlled, experimental variability can. Therefore, the variability of net nondimensional O_2 cost should be carefully examined, including the potential for protocol changes to improve reliability.

Statement of Clinical Significance

Improving reliability of O_2 cost measurement can improve its clinical utility.

Methods

A retrospective analysis was used to compare the variability in O_2 cost between two protocols (*old* and *new*) at a single center. Both protocols utilized the same breath-by-breath oxymeter (CPXD, Medical Graphics Corporation, St. Paul, MN).

Table 1: Comparison of Protocols

	Old Protocol	New Protocol
Rest Duration	3 minutes	10 minutes
Rest Steady State	Last minute of rest	Average of all 3 minute steady state breaths as determined by Kendall's tau
Walk Duration	6 minutes	6 minutes
Walk Steady State	3 minute period with smallest slope	Average of all 3 minute steady state breaths as determined by Kendall's tau

The subjects met the following criteria: *i*) diagnosis of CP, *ii*) no prior surgery, and *iii*) oxygen data collected in a barefoot condition. It was assumed that due to stability in demographics, clinicians, and referral patterns, these criteria would result in well matched samples. Steady state was identified using Kendall's tau-b [4]. Breath-by-breath oxygen consumption data was analyzed in three minute intervals. The breath closest to the center of the interval was assumed to be at steady state (H_o). Rejection of H_o over the interval at the .10 level using a two sided test indicated non-steady data (rising of falling trend). A second order polynomial was fit to the net-nondimensional cost vs. nondimensional speed data from each protocol. This fit provides the fewest degrees of freedom capable of matching the concave upward shape of O_2 cost for both typically developing subjects and those affected by CP.

Results

Matching of groups: Relevant clinical data showed that the old and new samples, while substantially different in size (N = 278 new, 995 old), were largely the same in terms of factors likely to impact oxygen cost. For example, distribution of the CP subtypes hemiplegia-diplegia-triplegia-quadriplegia were 12%-68%-13%-7% respectively for the new protocol and 13%-60%-16%-11% respectively for the old protocol. Birth weight, weeks of gestation, and weeks in the neonatal intensive care unit were statistically equal ($p>0.05$). The O_2 cost vs. nondimensional speed regressions showed essentially unchanged polynomial fits for the old and new data, suggesting that the samples were well matched, and the protocols were unbiased [Figure 1].

Reduction of variability: The mean square error (MSE) in the old protocol ($MSE_{old} = .262$) was substantially higher than that in the new ($MSE_{new} = .050$). This represents a ≈5x reduction in variability. The reported MSE reduction does not take into account that fact that 90 outliers with large errors were removed from the old protocol; therefore the "true" MSE reduction is even larger than reported.

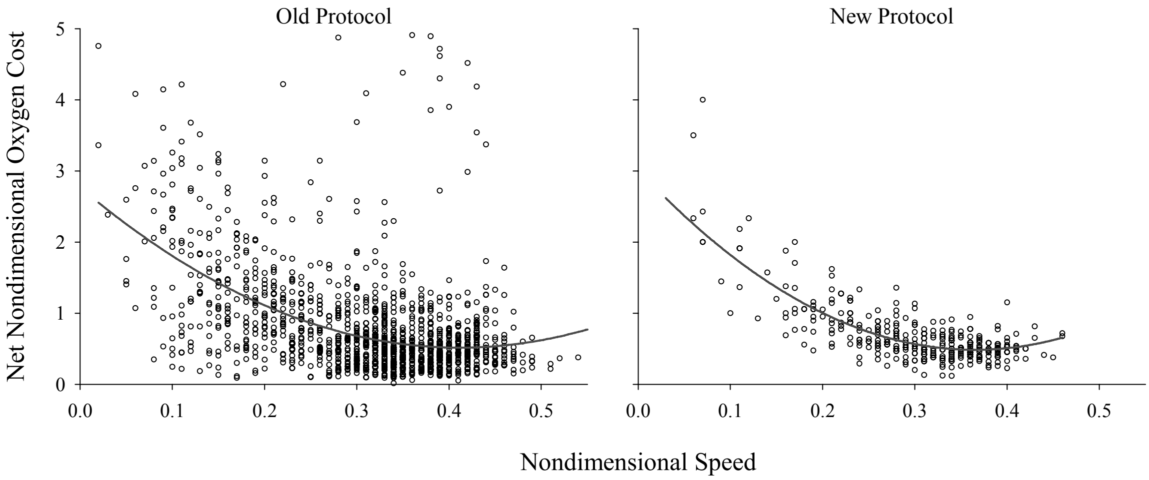

Figure 1*. Cost vs. speed plots show a reduction in variability with the new protocol. The quadratic regression is nearly identical suggesting that the protocols are unbiased and the groups are equivalent. Note that 90 extreme outliers (standardized residuals > 3) were eliminated from the regression fit for the old protocol – no such outliers existed for the new protocol. Thus the variability reduction reported is a conservative estimate.*

Discussion

A protocol utilizing a 10 minute resting period and a statistical determination of steady state substantially reduced the variability in O_2 cost data. Decreased variability is likely to lead to greater clinical utility, since smaller changes in O_2 cost can be reliably interpreted. Changes in both rest period (duration), and steady state determination (method) occurred. Thus it is not clear how much each factor influenced the improvement. While it is possible that the reduction in variability could be due to differences in samples, the groups appear to be well matched, except for age, where the subjects tested with the new protocol were significantly older than those tested with the old protocol (age = 8.3 yrs. new, 10.9 yrs. old).

References

1. Brehm MA and Harlaar J, Proc. 10th Ann. GCMAS, 2005
2. Schwartz MH *et al*., Gait Posture, in press.
3. Bowen TR *et al*., JPO, **18**:738-742, 1998.
4. Kendall MG, Time Series, New York, Hafner Press, 1973.

O-14

ALTERNATIVE METHODS FOR MEASURING TIBIAL TORSION

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Summary/conclusions
This study introduces several new techniques for determining tibial torsion using motion analysis data. Analysis of correlations suggests that the new techniques may be more clinically valuable than the traditional bi-malleolar axis measurement.

Introduction
Long bone torsions are notoriously difficult to measure [1]. The most common measure of tibial torsion is the bi-malleolar axis angle. Bony landmark ambiguities combine with line-of-sight problems resulting in an angle that is hard to accurately measure. Efforts to reduce the variability in several torsion measurements within the Gillette Children's Specialty Healthcare Center for Gait and Motion Analysis have been largely unsuccessful. This study uses motion analysis data to determine estimates of knee and ankle axes and thereby estimate tibial torsion. These "technical" measurements can be compared to the bi-malleolar axis angle.

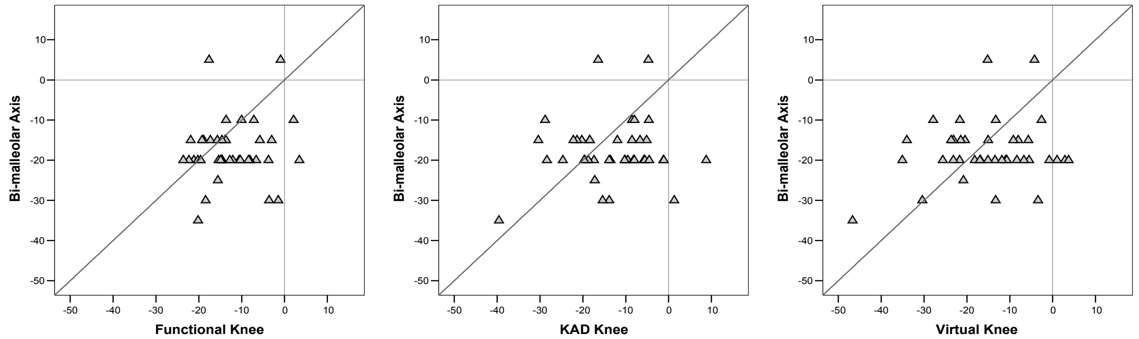
Statement of clinical significance
Better methods for measuring tibial torsion can lead to improved clinical decisions regarding tibial derotational osteotomies. Determination of tibial torsion using motion analysis data appears to be more useful than a bi-malleolar axis measurement.

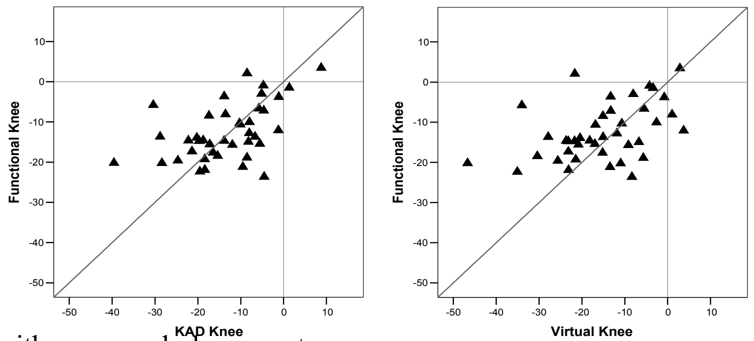
Methods
Data from 20 young, able-bodied volunteers was used in this study (ages 5-14 yrs). Four different tibial torsion measurements were acquired; one from a physical exam and three from motion analysis data. For the physical exam measure, the bi-malleolar axis angle was used. For the motion analysis methods the locations of the medial and lateral malleoli were estimated using virtual circles [2]. The centers of those virtual circles were then used to define a bi-malleolar axis. The knee axis was estimated using three different techniques. The first used the center of a virtual circle around the medial femoral condyle and a physical knee marker (*virtual knee axis*). The second was determined *via* placement of a knee alignment device (*KAD knee axis*). The third used a functional method (*functional knee axis*) described by Schwartz, *et al.* [3]. Note that the *functional knee axis* is defined independently from the *virtual* and *KAD knee axes*. For the tibial torsion measures calculated from the motion analysis data, the ankle and knee axes were projected onto a plane perpendicular to the long axis of the tibia. Tibial torsion was then taken to be the average of the angle between the projected knee and ankle axes during a static trial.

Results
The results show that there is no significant correlation ($p > 0.05$) between physical exam based tibial torsion and tibial torsion measured using the motion analysis data [Figure 1]. It is also shown that the three motion analysis based measurements are significantly correlated to each other ($p < 0.01$).

Discussion

The fact that physical exam based tibial torsion was uncorrelated with any of the other measures suggests one of two things: either the other three methods are wrong, or the physical exam measure is wrong. The fact that significant correlations ($p < 0.01$) existed between the *functional knee axis* and both the *virtual* and *KAD knee axes* suggests that the latter (physical exam wrong) is the case, since the mathematical definition of the *functional knee axis* is independent of the other two axis definitions. The highest correlation occurred between the *KAD* and the *virtual knee axes*, which is expected since the *KAD* and *virtual knee axes* share a common landmark (lateral femoral epicondyle).





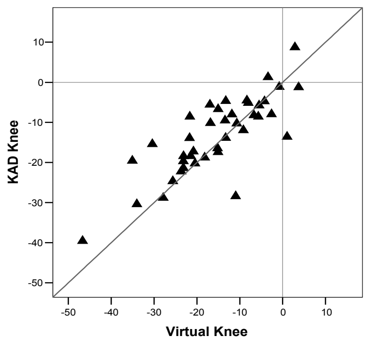


Figure 1. These scatter plots show comparisons between 4 different methods of measuring tibial torsion. All measurements are in degrees, with positive indicating an external tibial torsion. The diagonal lines have a slope of 1.
The measurements in the plots with grey symbols *are not* significantly correlated ($p > 0.05$), while measurements the plots with the black symbols *are* significantly correlated ($p < 0.01$). The significant correlations (r) are as follows: *functional* vs. *KAD* = 0.512, *functional* vs. *virtual* = 0.451, and *KAD* vs. *virtual* = 0.807.

References

[1] Schwartz MH and Sohrweide S, Proc. 13th annual ESMAC, Warsaw, Poland, 2004.
[2] Rozumalski A and Schwartz MH, Proc. 9th annual GCMAS, Lexington, USA, 2004.
[3] Schwartz MH and Rozumalski A, J. Biomech, Vol 38/1 pp 107-116, 2005.

O-15

OVERLAY PROJECTION OF 3D GAIT DATA ON CALIBRATED 2D VIDEO

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Summary/conclusions
A straightforward technique for the calibration of an arbitrarily placed video camera in the same laboratory coordinate system as motion capture cameras and force plates is described. Captured 3D data and derived data can then be projected onto recorded 2D video.

Introduction
Systems for video overlay of graphical analog data (eg EMG) or ground reaction force vectors (planar projections) have been developed in the past. However, despite the potential power of overlaying 3D information on video, projection of 3D data on 2D video recorded by an arbitrarily placed video camera has not been readily available. The straightforward calibration of a video camera in the same laboratory coordinate system as motion capture cameras and force plates in Matlab (Mathworks Inc, Massachusetts, USA) is described here. Captured 3D data is then projected onto recorded 2D video.

Statement of clinical significance
Overlay of 3D data on 2D video has many potential applications in clinical gait analysis for visualization / quality control of 3D reconstruction, labelling, virtual markers (eg joint centres), and muscle / joint modelling.

Methods
Video of gait trials is captured via Firewire to AVI file by Vicon Workstation (Vicon Peak, Oxford, UK) using a Fire-i camera (Unibrain SA, Athens, Greece), with 640x480 pixel resolution at 30 frames/sec (Figure 1). Eight Vicon M2 cameras (1280 x 940 at 120 frames/sec) are used for motion capture of the reflective markers. A fixed 'frame offset' is empirically determined to time-match the 30 fps video frames to the 120 fps motion capture data. To calibrate the Firewire video camera, multiple still images of a planar, 'checker-board' are separately captured in different orientations (Figure 2). The only information required about the planar board is the standard size of the squares. A second planar object is subsequently placed over one of the force plates (Figure 3) to establish the video camera position and orientation in relation to the laboratory coordinate system for motion capture.

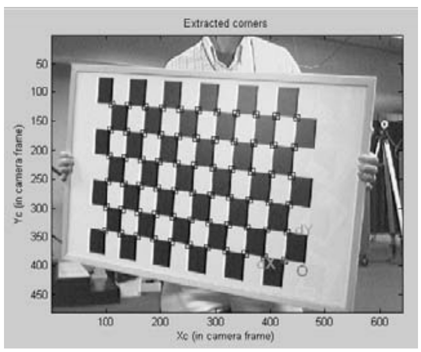

Figure 1: Unibrain Fire-i camera. **Figure 2**: Planar calibration board.

The still images are processed in the public domain Camera Calibration Toolbox for Matlab developed by Bouguet [1], based on methods exploiting the unique characteristics of planar calibration objects [2,3]. The intrinsic camera parameters (focal length, principal point, radial and tangential distortion) are initially determined from the multiple board views, followed by the extrinsic parameters (position and rotation matrix) from the board over the force platform. No information about the video camera characteristics needs to be supplied for its calibration. The intrinsic and extrinsic parameters of the Vicon M2 motion capture cameras are determined via the standard Vicon Workstation 'DynaCal' camera calibration routines, using a 238mm two-marker wand. In this example, the standard Vicon Plug-in-Gait lower limb marker set was used for the gait trial. After marker 3D reconstruction in Workstation, 3D data is retrieved from the recorded C3D file in Matlab using C3D Server (Motion Lab Systems, Louisiana, USA). Standard Matlab routines are used to read the video AVI frames in sequence, and a routine in the Camera Calibration Toolbox performs a projection of the time-matched 3D data onto the 2D video frames.

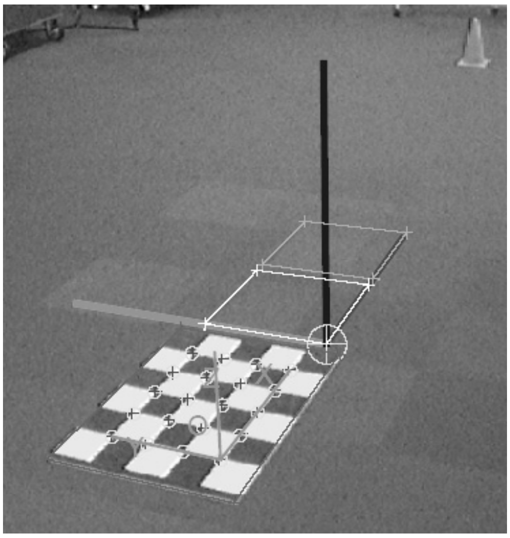

Figure 3: Extrinsic parameter determination.

Figure 4: Projection of 3D data onto video.

Results and Discussion

Figure 4 displays an example of the capabilities of the method showing, for simplicity, a single frame from the video with projections of the lateral knee marker trajectory and force plate locations. The presence of visible markers in the video sequence for which the 3D coordinates have been reconstructed provides a convenient in-built check that the projection is fundamentally accurate. In the case where a projected 3D coordinate of a given marker deviates slightly from the visual image of that reflective marker, it will be difficult to determine whether the small error lies in the reconstructed 3D coordinate or in the video camera calibration. However, more significant errors in 3D reconstruction, labelling, and virtual markers (eg joint centres) will be readily apparent, and much more obviously so than in standard stick figure or rendered skeleton representations without video overlay. Correct location and scaling of muscle and joint models will also be evident.

References

[1] Bouguet (2005) Camera Calibration Toolbox for Matlab. http://www.vision.caltech.edu/bouguetj/calib_doc/. Latest version August 10, 2005.

[2] Zhang (2000) A flexible new technique for camera calibration. IEEE Trans. Pattern Analysis & Machine Intelligence, 22(11):1330-1334.

[3] Heikkila & Silven (1997) A four-step camera calibration procedure with implicit image correction. Procs IEEE CVPR97, pp 1106-1112.

O-16

RELATIONSHIP BETWEEN THE ANTHROPOMETRIC VARIABLES AND FRONTAL KNEE MOMENTS IN HEALTHY OBESE ADULTS.

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Summary/conclusions
Increased body weight has been proposed as a risk factor for the development of knee OA. The relationship between body mass and the frontal plane knee moment, however, is only moderate. The knee moment is also influenced by individual anthropometric characteristics that help to define the distribution of mass.

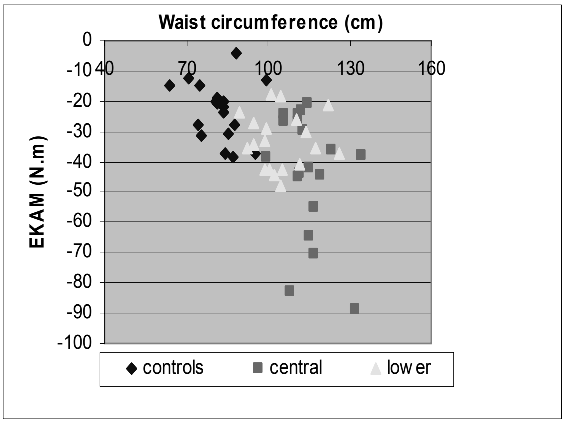
Introduction
Osteoarthritis (OA) is a leading cause of disability among US adults causing pain, joint stiffness and limited mobility[1]. Obesity has been identified as an important modifiable risk factor contributing to both the development [2] and progression of knee OA [3]. Researchers believe that although obesity may be affecting the pathogenesis of knee OA through both biochemical and biomechanical pathways, mechanical loading of the articular cartilage is of primary importance. Obese women have an increased prevalence of knee OA compared to men [4]. In addition, gender differences in obesity patterns and other anthropometric variables have been documented. This raises the possibility of associations between body anthropometrics and altered gait biomechanics that put certain obese adults at an increased risk of developing knee OA [5].The external knee adduction moment (EKAM), thought to be a contributor to medial compartment knee OA [6], provides a good estimate of knee joint loading. Correlating EKAM with anthropometric variables, such as waist, hip, thigh, gluteal furrow circumferences of obese adults could help to uncover associations that might be useful in identifying obese populations at risk of developing knee OA.

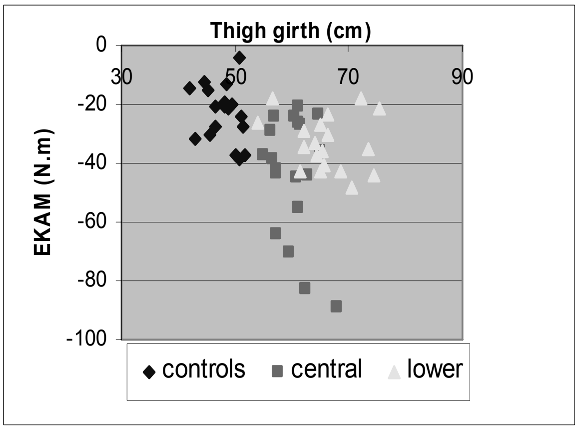
Statement of clinical significance
Over 70% of US adults are either overweight or obese and thus at a higher risk for developing knee OA [7]. Trying to identify issues that affect loading at the knee in this population may help to identify intervention strategies that can be used to affect the incidence of OA in this population.

Methods
Sixty subjects, twenty normal weight controls, twenty subjects with a lower obese pattern and twenty subjects with a central obesity pattern, between the ages of 35-55 years and with no knee, lower limb, neuromuscular or medical problems participated in the study. Obesity was defined as having a body mass index (BMI) greater than or equal to 30.0 Kg/m^2 (BMI of controls < 25.0 Kg/m^2). Women with waist:hip (W:H) ratio of 0.85 or less and men with a ratio of 0.95 or less were identified to have lower obesity patterns. Anthropometric variables of height, weight and waist, hip, mid-thigh, gluteal furrow circumferences were measured as per a set protocol. Standard kinetic and kinematic data was collected using an Optotrak motion analysis system and Kistler force plate as the subjects walked along a 10 m walkway at their self selected speed. Three non-collinear markers were used to track the right lower limb, pelvis and trunk. Kinematic data were collected at 60 Hz and filtered at 6 Hz. Frontal plane data were analyzed using Visual 3D (C-Motion, Inc) to obtain the external knee adduction moments. Inter-group comparisons were made using a One-way ANOVA and Pearson's correlation coefficients were calculated to assess associations.

Result

A significant difference in the means of the peak EKAM was found in the central obese pattern and the controls group ($p<0.001$) whereas there was no difference between obese groups ($p=0.82$). The associations (r) between anthropometric variables and EKAM were found to be both moderate: waist circumference (r=-0.52), hip circumference (r=-0.48), BMI (r=-0.43); and weak: thigh girth (r=-0.38), gluteal furrow girth (r=-0.36). Additionally, a step-wise regression procedure defined a best fit model consisting of BMI, waist circumference, thigh girth and W: thigh ratio that could explain 36.7% of variance (r^2) found in the knee moments.





Discussion

The results of the study demonstrate that certain anthropometric measures are associated with the EKAM. While three of the four anthropometric variables that had the strongest associations might be considered as being more closely related to a lower obesity pattern, when subjects were divided based on the W:H ratio (as per the original hypothesis) obesity groups were not found to be significantly different. It is also worthwhile to note that in spite of their increased weight, compensatory gait strategies developed by these healthy obese adults helped to reduce the adverse knee joint loads and these compensations should also be looked out for. Further exploration of how these variables interact to affect the EKAM might be helpful in considering appropriate interventions, although the strength of these relationships will require that any interpretations be conservative.

References

1. Felson, D.T., *Epidemiology of Osteoarthritis* K. Brandt, et al Editors. 1998, Oxford U Press:. p. 12-22.
2. Gelber, A.C., et al., Am J Med, 1999. **107**(6): p. 542-8.
3. Spector, T.D., et al,. Ann Rheum Dis, 1994. **53**(9): p. 565-8.
4. Manninen, P., et al., Int J Obes Relat Metab Disord, 1996. **20**(6): p. 595-7.
5. Radin, E.L., et al. Lancet, 1972. **1**(7749): p. 519-22.
6. Baliunas, A.J., et al., Osteoarthritis Cartilage, 2002. **10**(7): p. 573-9.
7. Patterson, R.E., et al., Am J Prev Med, 2004. **27**(5): p. 385-90.

O-17

COMPARING THE RELIABILITY OF VIRTUAL GONIOMETRY AND UNIVERSAL GONIOMETRY

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Summary/Conclusions

This study investigated the reliability of virtual goniometry (VG) and the universal goniometer (UG) for the measurement of sagittal angles at the elbow and knee. VG measurements were made by taking digital photographs of static joint angles, transferring the images to a PC and finally manipulated computer software's virtual measurement arms (VMA) to calculate the joint angle. Three different examiners were used and both intra-rater and inter-rate reliability was investigated. Four different approaches to the alignment of the VMAs over the joint images were tested. Alignment of the VMAs along anterior-posterior bisection points of the limb segments provided the strongest reliability. The use of adhesive markers at anatomical landmarks did not improve reliability. The standard UG was shown to have poor reliability scores compared with VG.

Introduction

The aspiration for a scientific approach to the management of neuromuscular and orthopaedic disease has a fundamental requirement for accurate measurement tools. Reliable tools that allow the clinician to monitor joint range of movement and function and to determine the effect of particular interventions are required. The literature has not shown the standard UG as having strong reliability and VG tools such as the Uillinn Method© [1] or the Internet Based Goniometer [2] have been proposed as an alternative to UG. A comparison between the reliability scores for the VG and UG would help determine optimal measurement practice.

Statement of clinical significance

This study presented data on the reliability of VG and compared it to the reliability of the standard UG. The results provide clinically relevant information regarding the anticipated error associated with routine static joint angle measurement and suggests ways to improve measurement accuracy.

Methods

Measurements of sagittal joint angles at the knee and elbow were used in this study. The VG used in this study was the Uillinn Method© [1]. The 4 different approaches to applying the VMAs about the joints were; the *No Marker* method which had the examiners place the VMAs where they perceived the standard anatomical landmarks to be (skin markers were not used), the *Parallel* method placed the VMAs over the approximated longitudinal axis of the limb segments, the *Mid-Point* method used additional software markers to identify the anterior-posterior bisection at different points along the limb segments. These were then used as a guide to placing the VMAs. Finally, the *Skin Marker* method used 3 small adhesive markers placed at anatomical landmarks upon which to align the VMAs. Each of the subjects involved in the testing were required to remain stationary for 15 minutes while 3 chartered physiotherapists repeatedly measured and photographed the 51 individual knees and 49 elbows that were placed in random positions. Healthy subjects were chosen to minimise biological error. Data was gathered for both intra-rater and inter-rater reliability for both the UG and each of the 4 VMA placement methods. Blinded precautions were observed. The agreement statistics used were the Repeatability coefficient (RC), Limits of Agreement (LOA) and Typical Error (TE) which provided clinically relevant data. Correlation coefficients (ICC) were also used to complete the analysis.

Results

A limit of systematic bias was set at 2.1°, this figure was based on the average magnitude of the angles tested. None of the measurement techniques had an intra-rater systematic bias. Only the No Marker (elbow), the Parallel (elbow) and the UG (elbow) had a systematic bias between the different examiners.

Table 1. Intra-rater and Inter-rater Reliability: Typical Error for all Examiners.

	Intra-rater Reliability						Inter-rater Reliability					
	Knee			Elbow			Knee			Elbow		
	RC	TE	ICC	RC	TE	ICC	LOA	TE	ICC	LOA	TE	ICC
Universal Goniometer	5.8°	2.1°	0.99	6.9°	2.4°	0.99	-10.6°, 10.3°	3.99°	0.89	-11.1°, 8.1°	7.19°	0.90
UM (**No Markers**)	3.7°	1.3°	0.99	4.8°	1.9°	0.99	-5.5°, 5.7°	2.44°	0.97	-7.5°, 4.9°	3°	0.93
UM (**Parallel**)	2.9°	1.1°	0.99	4.1°	1.8°	0.99	-7.2°, 7.9°	2.8°	0.95	-10°, 6°	3.4°	0.90
UM (**Mid-Point**)	2.8°	1°	0.99	3.5°	1.6°	0.99	-5°, 5.7°	1.99°	0.97	-3.4°, 5°	2.1°	0.95
UM (**Skin Markers**)	2.9°	1.1°	0.99	4.0°	1.9°	0.99	-3.3°, 5.3°	1.75°	0.98	-6.2°, 5.9°	2.16°	0.94

UM=Uillinn Method© : ICC= Intraclass correlation coefficient (3,1), LOA=Limits of agreement: RC=Repeatability coefficient

Discussion

Measurement error arises from one of three different sources: equipment error, examiner error or biological error [3]. A previous related study established the equipment error associated with the Uillinn Method© of VG to be typically (TE) less than 0.3°, which is negligible. As this study controlled biological error, our results largely reflect error attributable to the examiner. Indeed there were some differences between the scores of the individual examiners suggesting different levels of skill. VG has the ability to control some variables that the UG cannot. For example repeated placement of the measurement arms to re-assess data subsequent to patient involvement is only possible with the VG due to the recorded photograph of the joint. Examiner interaction with the limb is limited compared to UG, which may lessen the influence of examiner bias. It seems that the UG is more reliant on examiner skill than the VG. Our results suggest that examiner error may be a substantial factor to reliability, which may explain why the VG results were stronger than those for the UG.
The lack of sensitive and reliable tools for the clinical measurement of joint angles is a major issue for health science. The continued development of a scientific approach to managing movement disorders requires that such tools be developed and properly tested.

References

[1] Dunlevy, C., Cooney, M., Gormley J. (2005) Physiotherapy Research International, 10(4) 190-200
[2] Russell, TG., Jull, GA., Wootton, R., (2003) Can the Internet be used as a medium to evaluate knee angle? Manual Therapy. 8(4) 242-246.
[3] Bruton, A., Ellis, B., Goddard, J. (1999) Comparison of Visual Estimation and Goniometry for Assessment of the Metacarpophalangeal Joint Angle. Physiotherapy. 85(4) 201-208.

O-18

3D HIP AND PELVIC GAIT PATTERN RECOGNITION IN CHILDREN WITH CP

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Summary/conclusions
Cluster analysis was used to recognize typical patterns and to search for underlying pathological mechanisms which might explain the large inter-subject variability in hip and pelvic motion during gait in children with CP. Five typical gait patterns were clinically described. Two groups of older children presented with continuous internal hip rotation (one group with flexed hip and another with full hip extension at terminal stance), and two groups of young children with gradually increasing hip rotation during stance (also one group with flexed hip and another with full hip extension at terminal stance). A final group of mainly children with hemiplegia only showed mild hip problems. The 5 clusters were characterised by significant between-group differences for pelvic and hip gait parameters and for clinical measures of muscle contractures and strength, but not for clinical alignment measures.

Introduction
Children with cerebral palsy (CP) present with a variety of gait patterns and in general, there is a low correlation between clinical parameters (ROM, alignment, spasticity , strength and selectivity) and gait deviations [1]. Therefore, the underlying mechanisms of the pathological gait are still poorly understood. Gait classification systems have been developed to structure and to create more insight into different gait patterns. However, most gait classifications only describe sagittal gait patterns, thereby mainly focusing on ankle, knee and sagittal hip motion. However, 3D hip and pelvic pathology is frequently observed in children with CP, and is known to be complex, influenced by a combination of bony deformities, increased tone, muscle contractures and weakness [2]. The purpose of the study was to evaluate whether different gait patterns could be statistically recognized and clinically described for 3D hip and pelvic motion for a large group of ambulant children with CP, by using cluster analysis. It was hypothesised that the specific gait patterns would be related to diagnosis, age and clinical measurements (alignment, ROM, tone, strength and selectivity).

Statement of clinical significance
More insight into the typical 3D hip and pelvic gait patterns can help to understand the underlying mechanisms of pathological gait, and eventually improve communication between clinicians and facilitate clinical decision-making.

Methods
342 ambulant patients (189 with diplegia and 153 with hemiplegia) were selected for this retrospective study. The inclusion criteria were: (1) ambulation, without walking aids, (2) full gait analysis, including 3D kinematics, kinetics and EMG (VICON, AMTI, K-lab) and full clinical examination (ROM, spasticity, strength and selectivity) at 4 to 20 years of age. 40 gait parameters (average of 3 trials per subject) and 32 clinical examination parameters were defined for each subject. To avoid the influence of inter-correlation between limbs, only one pathological side per subject was included for further analysis. Non-hierarchical k-means clustering was used on the standardised data of a selection of 18 kinematic and kinetic parameters of hip and pelvis, resulting in 5 clusters. The definition of the number of cluster was based on the study of intra and inter cluster variability and the frequency distributions of the clusters. The 5 clusters were clinically described by comparing the gait parameters with the parameters of an age-related normal control group of 68 children. Mean and SD of all

continuous parameters and median and interquartile range, as well as frequency distributions for categorical variables were calculated for each of the 5 clusters. Analysis of variance (ANOVA, post-hoc Tuckey and Kruskal-Wallis analysis) was used to detect differences in gait and clinical examination parameters between the 5 clusters.

Results and discussion

We found five clinically meaningful groups of gait patterns (Fig. 1). The most discriminating parameters were found in hip rotation, sagittal hip motion and pelvic tilt. Two groups of children showed a clear **continuous** internal hip rotation, slightly increasing during stance (blue and pink graph in Fig. 1). This internal hip rotation was significantly higher compared to the other groups and to the normal controls (p<0.0001). However, both groups were further discriminated because of a different hip flexion/extension pattern, with one group (N=51) fixed in flexed position, with significantly decreased sagittal hip ROM (P<0.0001) (i.e. *continuous flexed internal hip* = blue graph), and the another group (N=56) presenting with a full hip extension at terminal stance (i.e. *continuous pure internal hip* = pink graph). Both groups had increase anterior tilt, with increased sagittal pelvic ROM (significantly increased compared to the three other groups and compared to normal controls, P<0.005), and reduced hip abduction moment (P<0.0001). The pattern of continuous flexed internal hip was further characterised by significantly increased hip adduction in stance, retracted pelvis, delayed and reduced hip flexion moment and reduced H3 power burst. Two other groups of children showed a typical **dynamic** internal hip rotation, characterised by normal hip rotation at initial contact, patholgically increasing during stance (green and black graph in Fig. 1). Related to the dynamic hip rotation, also ROM for pelvic rotation was significantly higher (P<0.0001) compared to the other groups and to normals. Again, both groups were further discriminated because of a different hip flexion/extension pattern, with one group (N=49) predominantly in flexed position, with significantly decreased sagittal hip ROM (P<0.0001) (i.e. *dynamic flexed internal hip* = green graph), and the another group (N=61) presenting with a full hip extension at terminal stance (i.e. *dynamic pure internal hip* = black graph). Children with dynamic pure internal hip showed the largest increased anterior tilt combined with increased sagittal pelvic ROM (typical double bump pattern). Just like for the continuous flexed internal hip, the pattern of dynamic flexed internal hip was further characterised by significantly delayed and reduced hip flexion moment (P<0.0001). A final group of 125 children (red graph) only showed *mild hip and pelvic problems* in gait. Hip rotation, as well as sagittal and frontal hip motions were all within normal ranges. Frequency distributions for age and diagnosis among clusters revealed that the mild pattern was most frequently observed in young children with hemiplegia. 48.4% of the hemiplegic children presented with these mild hip/pelvic problems, compared to 27.0% of the diplegic children. Children with hemiplegia in this study seldom walk with continuous pure internal hip (only 5.2%). From a detailed study of the clinical parameters for the 5 clusters it could be concluded that clusters were mainly discriminated by differences in muscle contractures (hip adductors and hamstrings), strength of hip abductors and hamstrings spasticity. The 5 clusters showed similar clinical measures for alignment (f.i. femoral anteversion).

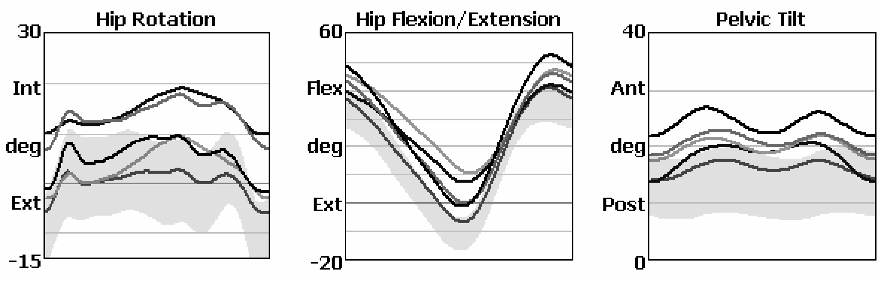

Figure 1: hip and pelvic patterns for 5 clusters

References

[1] Desloovere et al.(2006) Do dynamic and static clinical measures correlate …, *Gait&Posture* (in press);

[2] Kerr et al., The midpoint of passive hip rotat …, 2003, Gait&Posture 17:88-91.

O-19

RELIABILITY AND VALIDITY OF AN ACTIVITY MONITOR (IDEEA®) IN A PAEDIATRIC POPULATION.

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Summary / conclusions

This pilot study demonstrated that the IDEEA® activity monitor has moderate to high levels of reliability detecting everyday activities in a normal paediatric population. Concerns were highlighted in the accurate detection of walking and gait parameters such as, step length and cadence which need to be addressed prior to use in a pathological patient population.

Introduction

Three-dimensional gait analysis (3-DGA) provides excellent quantitative, reliable measures of gait abnormalities and is used to determine and monitor orthopaedic surgical interventions in children with cerebral palsy. However, the gait analysis lab setting is an artificial environment and it is unknown how changes measured by 3-DGA reflect in the child's community performance. Recently a light-weight, wearable device (IDEEA®, MiniSun LLC[1]) has shown high reliability in adults in quantifying community walking and levels of daily activity[2]. This portable system allows the assessment of both temporo-spatial gait parameters and a range of functional activities, using a 5 triaxial accelerometer system[1]. As yet, there is no information regarding the function of this IDEEA® device in the paediatric population. The aim of this paper is to i) present preliminary reliability and validity findings on a comparison of gait parameters determined from the IDEEA® to those collected simultaneously during a 3-DGA and ii) determine the level of accuracy of the IDEEA® in identifying activities in a group of control children. Future studies aim to examine the reliability of this device in children with cerebral palsy in the community setting.

Statement of clinical significance

In accordance with World Health Organisation guidelines, clinicians are encouraged to assess clinical outcome in terms of changes in the person's activity levels and participation[3]. The IDEEA® could be used to monitor the child's functional ability in the community following surgical and rehabilitation interventions.

Methods

Ethical approval was obtained for 12 control children. Part 1 of the study compared the gait parameters of velocity (metres/second), stride length (metres), step length (metres) and cadence (steps/minute) obtained from the IDEEA® to those simultaneously collected from a 3-DGA. Part 2 determined the accuracy of the IDEEA® in detecting 5 activities including: walking, stairs, standing, sitting and lying. The IDEEA® consists of a small box clipped to the waistband with 5 sensors attached to the legs, feet and trunk. Once attached the IDEEA® is calibrated before each measurement session. A stopwatch was started as soon as the IDEEA® recording began to allow precise matching to 3-DGA and recorded activities. The children first completed a 3-DGA with both the IDEEA® on and 21 retro-reflective markers required for 3-DGA. Children completed a 10 metre walk six times, with data collected on a Vicon Workstation (version 5.0) at 60HZ. Subjects then completed a set protocol of 4 activities, progressing from a position of supine lying to sitting and then standing, maintaining each position for 30 seconds. Subjects then walked up and down a flight of stairs. The identification of walking activity was obtained from the previous 3-DGA walks. Each child completed two identical data collections, one week apart at the University of Auckland Gait Laboratory. The percentage accuracy of the IDEEA® in determining each of the 5 activities was calculated by comparing the recorded time that the subject performed each activity with the IDEEA® output

(MiniSun LLC, GaitView software).The IDEEA® is documented to identify a multitude of different activities. For our purposes, these were grouped into 5 main activities – lying, sitting, standing, walking and stairs. The gait parameters measured by 3-DGA and IDEEA® were compared using Bland Altman limits of agreement, which graph the mean of the two measures against the difference between the two measures[4], plus 95% limits of agreement (± 2SD) combining session1 and session 2 data.

Results

12 control subjects consented (mean age 11 years, range 7-21 years; 6 male: 6 female). Figure 1 shows the difference between the IDEEA® and 3-DGA obtained measures of walking velocity, with the 95% limits of agreement of ± 0.3 m/sec. Larger 95% limits of agreement between the measures were found for stride length (0.6 to -0.08 m), step length (0.35 to -0.05m) and cadence (2 to -21steps/min) over identical walking conditions. The IDEEA® overestimated step length (with a bias difference of +0.15m) and underestimated cadence (with a bias difference of -10 steps/min). Figure 2 shows the identification of activities by the IDEEA® over both sessions. The IDEEA® identified lying and standing 100% and 99% of the time respectively. Walking was only identified 72% of the time, with running being detected 15% or stairs 5% of the time when children walked at their self selected speed. Sitting was detected as either sitting or lying up to 90% of the time. We do not know how the IDEEA® defines sitting vs. reclining vs. lying, which may account for this ambiguous result.

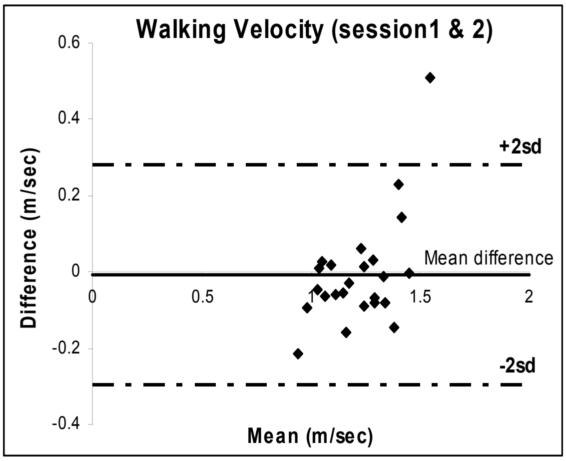

Figure 1. Difference between IDEEA and 3-DGA measures versus the mean walking velocity

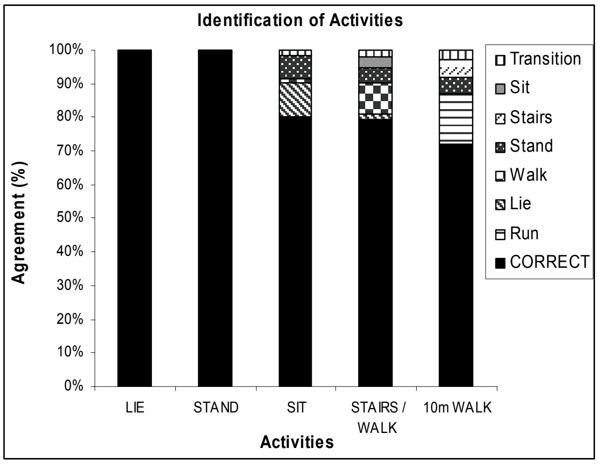

Figure 2. Identification of activities by IDEEA®

Discussion

The aim of this pilot study was to assess the accuracy of the IDEEA® activity monitor in determining everyday movements and gait parameters in a paediatric population. To date, the IDEEA® has been found to have moderate to high levels of accuracy in determining selected activities. Lying and standing are very accurately recorded, similar to studies in adult populations. However, importantly walking at a self selected speed was detected as running or walking up stairs for 20% of the time. Comparison of measured gait parameters between the IDEEA® and simultaneous 3-DGA found a moderate agreement for walking velocity, but bias in measures of step length and cadence.

References

1. IDEEA®, Mini Sun LLC, USA; 2. Zhang et al. (2003), Obesity Research, 11, 33-40; 3. ICF, (2001), World Health Organisation; 4.Bland et al (1986), Lancet, 1, 307-10

O-20

HOW ACCURATE IS THE ESTIMATION OF ELBOW KINEMATICS USING ISB RECOMMENDED JOINT COORDINATE SYSTEMS?

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Summary/conclusions
The International Society of Biomechanics (ISB) proposes the use of standardized Joint Coordinate Systems (JCS) for the description of elbow kinematics. The use of JCS imposes simplification over the position and orientation of the elbow flexion-extension and prono-supination axes. The aim of this work was to access the effects of these assumptions by comparing the kinematics estimated by ISB suggested JCSs with respect to a known motion imposed to the hinge joints of the two reference models presented in [1]. Results showed that during a pure flexion-extension movement of 140° of range, JCSs estimate a fictitious prono-supination ranging from 12° to 16°. These results suggest a careful interpretation of in-vivo kinematic data, in particular those related to complex activities of the daily living.

Introduction
The ISB has recently realised recommendations for the description of elbow kinematics [2], based on JCS. These require the definition of a bone-embedded reference frame for each bone forming the joint, that is humerus and forearm. Elbow axes of rotations are assumed, therefore, as the medio-lateral axis of the humerus frame (flexion-extension) and as the caudo-cranial axis of the forearm (prono-supination). Since these are conventionally defined based on anatomical landmarks, they are only approximations of the real joint rotation axes. The aim of this work was to quantify the errors in the estimation of elbow kinematics when using ISB recommended JCSs, compared to a known motion imposed to the hinge joints of two elbow reference models s2R and s4R, derived from cadavers data [1].

Statement of clinical significance
The assessment of the accuracy of JCSs estimated kinematics is essential for a correct interpretation of in-vivo motion analysis results in clinics.

Methods
The ISB suggests the use of two alternative bone-embedded frames for the humerus, that is H1 and H2, and the system of reference F for the forearm [2]. It is important here to notice that, while H1 medio-lateral axis is only based on humerus anatomical landmarks (ALs), that of H2 is defined as normal to the long axis of humerus and forearm when the elbow is flexed 90° and completely supinated (reference pose). This means that for H2 definition the coordinates of US and RS in the reference pose must be known. From these systems of reference, the JCSs matrices ${}^{H1}R_F$ and ${}^{H2}R_F$ are defined. The decomposition of these matrices with the Euler sequence Z-X'-Y'' provides the estimation of the elbow flexion-extension (FE), carrying angle (CA) and prono-supination (PS) patterns. In order to assess the errors coming from JCS application, a pure, single elbow flexion-extension movement of 140° of range was applied to the models s2R and s4R, while keeping a full pronation, starting from the anatomical posture. Before computing ${}^{H1}R_F$ and ${}^{H2}R_F$ values during the motion (this was made possible since the models embed all the necessary ALs), H2 was defined by obtaining US and RS coordinates in H2 reference pose: a maximum supination of 175° was assumed for the cadavers [3]. In addition to the kinematics from ${}^{H1}R_F$ and ${}^{H2}R_F$, we also computed the relative orientation of the models Denavit-Hartenberg (D-H) systems of reference ${}^{G}T_{L1}$ and ${}^{G}T_{L3}$, obtaining ${}^{L1}T_{L3}$ over time. Thanks to D-H definitions, the decomposition of ${}^{L1}T_{L3}$ with the Euler sequence Z-X'-Z'' returns the gold standard estimation of

elbow kinematics, with the Z and Z'' patterns identical to those applied to the models hinge joints θ_{L2} and θ_{L3} and X' measuring the CA. The FE, PS and CA ranges of motion estimated by ${}^{H1}R_F$ and ${}^{H2}R_F$ were compared to those reported by ${}^{L1}T_{L3}$: the differences were considered the measure of the JCSs kinematic error.

Results

Table 1 reports the s2R and s4R ranges of motion for FE, CA and PS from ${}^{H1}R_F$ and ${}^{H2}R_F$ ${}^{L1}T_{L3}$, while figure 1 reports CA and PS motion patterns. CA patterns were reported negated to be consistent with data in the literature [4].

Discussion

While ${}^{H1}R_F$ and ${}^{H2}R_F$ estimation of FE differs slightly from the imposed motion (less then 5°), the greatest difference appear for CA and PS. Considering CA, s2R and s4R show different patterns which are consistent with the literature for the range of variation [4] and representative of known inter-subject variability [3]. From a robotics viewpoint, however, the only correct value of CA is that of ${}^{L1}T_{L3}$, since the CA is an intrinsic, geometric parameter of the robots, that cannot change in time. Considering PS, ${}^{H1}R_F$ and ${}^{H2}R_F$ reported a range of 12.16° and 15.56°, respectively, instead of the constant value obtained from ${}^{L1}T_{L3}$. These differences cannot be drawn back to a difference in definition (as for CA), but are due to the approximated definition of elbow axes. The reported errors are of great magnitude and suggest a careful interpretation of in-vivo kinematic data, in particular those related to complex activities of the daily living. Simulations with different levels of PS are currently underway.

	s2R			s4R		
	${}^{L1}T_{L3}$	${}^{H1}R_F$	${}^{H2}R_F$	${}^{L1}T_{L3}$	${}^{H1}R_F$	${}^{H2}R_F$
FLEX	140,00	135,28	135,26	140,00	135,47	135,04
CA	0,00	16,01	17,39	0,00	9,26	11,62
PS	0,00	15,72	12,16	0,00	11,43	15,56

Table 1. Ranges of motion in degrees

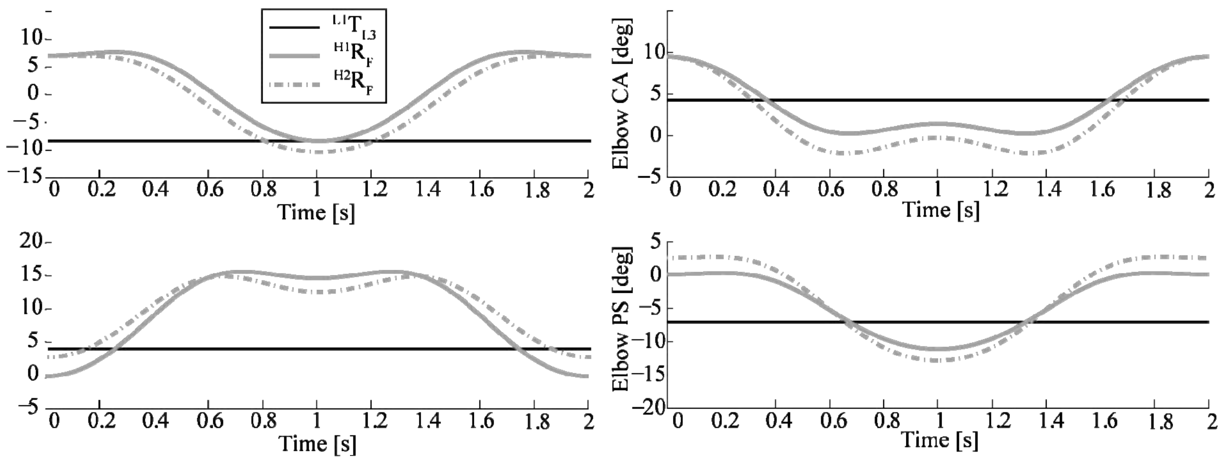

Figure 1. CA and PS patterns for s2R (left) and s4R (right)

References

[1] Cutti, A. G. et al., (2006), JEGM06 proceedings, *submitted*

[2] Wu, G. et al., (2005), J. Biomech., 38, 981-992

[3] Kapandji, I. A. (1982), Churchill Livingstone

[4] Goto, A. et al., (2004), J. Shoulder. Elbow. Surg., 13, 441-447

O-21

IMPACT OF WHEELCHAIR PROPULSION BIOMECHANICS ON DEVELOPMENT OF SHOULDER PAIN IN INDIVIDUALS WITH SPINAL CORD INJURY

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Summary/conclusions
Individuals with spinal cord injury (SCI) who developed shoulder joint pain over a 10 to 12 year follow-up period had a functionally inefficient pattern of WCP at the initial evaluation prior to the onset of shoulder pain than those who remained pain-free.

Introduction
The prevalence of shoulder joint pain and pathology is significantly higher in individuals after SCI than in the able-bodied population.[1] The most common diagnoses for those with SCI who have shoulder pain are rotator cuff impingement and tears which have been related to increased upper extremity weight bearing during wheelchair propulsion (WCP), transfers and raises.[2] The purpose of our retrospective study was to determine if an individual's biomechanical pattern of WCP could predict eventual development of shoulder joint pain. We hypothesized that the vertical component of the shoulder joint reaction force during WCP would be significantly greater in subjects who eventually developed shoulder joint pain. In addition, we wanted to determine which of the stroke parameters was correlated with the vertical shoulder joint reaction force to identify variables that could be altered to reduce vertical shoulder forces during WCP.

Statement of clinical significance
If some individuals with SCI have a WCP pattern that is associated with the development of shoulder pain, a training program to produce a propulsion pattern that would decrease the vertical forces at the shoulder and potential for impingement would be appropriate.

Methods
We attempted to contact 39 past participants in our previous research on WCP conducted in the early 1990's.[3,4] We were able to interview 12 individuals by telephone (11 men and 1 woman). While all subjects were free of shoulder joint pain at the time of the original study, 7 of 12 (58%) now reported the development of shoulder joint pain that was significant enough to seek medical treatment. We compared subject demographics and biomechanical parameters of WCP from the prior test for the 11 male subjects (6 with pain, 5 asymptomatic) using a two-group t-test. WCP variables included propulsion characteristics, push-rim reaction forces from an instrumented pushrim, arc and location of hand contact on the pushrim, shoulder joint reaction forces, and intensity and duration of electromyographic (EMG) activity of selected shoulder and scapular muscles. Given the small sample size, we considered any comparison with a p value of less than 0.15 a significant difference.

Results
There were no differences in age, years post SCI, or body weight between subjects who developed shoulder pain and those who remained pain free. Several trends were identified in the biomechanical variables that differentiated the shoulder pain from the pain-free subjects (Tables 1 and 2). The vertical component of the shoulder joint reaction force was greater in free WCP for the subjects who eventually developed pain. Moreover, cadence was higher in those who developed shoulder pain than in the pain-free group while velocity and cycle length were statistically similar. The pattern of force application on the push rim also was different in the two groups. Subjects who later developed shoulder pain had a smaller push arc with a

more forward initial hand placement and a greater vertical force on the push rim. Peak vertical shoulder joint force was moderately correlated with cadence (r = 0.58) and negatively correlated with push arc (r = -0.56). Intensity and duration of pectoralis major, intensity of serratus anterior and duration of supraspinatus and anterior deltoid activity were greater in those with pain.

Table 1. **WCP characteristics of subjects with paraplegia asymptomatic and painful shoulders**

	No Pain (n=5)	Painful (n=6)
FREE		
Velocity (m/min)	86.9 +/- 15	99.9 +/- 23#
Cadence (cycles/min)	**63 +/- 12**	**80 +/- 19 †**
Cycle Length (meters)	1.40+/- 0.15	1.33 +/- 0.32#
Vertical Shoulder Force (N)	**8.0 +/- 11**	**21.3 +/- 17 †**
Push arc	**66 +/-7***	**52 +/-7 ***
Angle at initial hand contact	**-24° +/- 6***	**-17° +/- 3 ***
Vertical Fz force on push rim (N)	**36+/- 4***	**58+/- 14***

Table 2. **Muscle activity of subjects with paraplegia with asymptomatic and painful shoulders**

	No Pain (n=5)	Painful (n=6)
Pectoralis Major intensity (%max)	**24 +/- 9**	**41 +/- 15 ***
Pectoralis Major duration (%cycle)	**25 +/-7**	**35 +/-6 ***
Serratus Anterior intensity (%max)	**13 +/- 12**	**34 +/- 26 †**
Supraspinatus duration (%cycle)	**52 +/-22**	**68 +/-17 †**
Anterior Deltoid duration (%cycle)	**25 +/-3**	**33 +/-5 †**

*** denotes p<0.05; † denotes p<0.15; # denotes p>0.35**

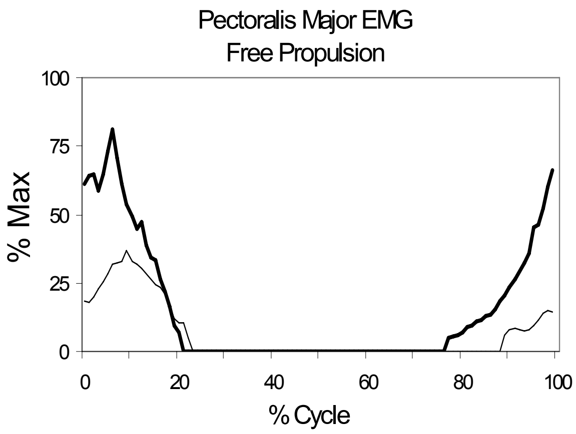

Figure 1. Mean sternal pectoralis major EMG during FREE propulsion. Thick line=painful group; Thin line=asymptomatic group

Discussion

Individuals who developed shoulder joint pain over 10 to 12 years had a functionally inefficient pattern of WCP at the initial evaluation than those who remained pain-free. Despite a similar velocity, those who developed pain had increases in cadence, vertical shoulder joint force and intensity and duration of EMG in the primary push phase muscles at the shoulder. These results combine to paint a picture of both higher vertical shoulder forces and greater potential for fatigue in the shoulder muscles. Individuals with weakness in these muscle groups would have reduced capacity to respond to the high shoulder forces and protect the glenohumeral joint during prolonged WCP. These results support the recommendations to use a slower cadence and a longer and more posterior arc of contact with the pushrim to decrease the vertical forces on the shoulder and the potential for impingement.[5]

References

1. Sie, IH, Waters, RL, Adkins, RH, and Gellman, H: Arch. Phys. Med. Rehabil. 73:44-48, 1992.
2. Escobedo, EM, Hunter, JC, Hollister, MC, Patten, RM, and Goldstein, B:Am. J. Roent. 168:919-923, 1997.
3. Kulig, K, Rao, S, Mulroy, S, Newsam, C, Gronley, J, Bontrager, E, and Perry, J: Clin. Orthop. Rel. Res. 354:132-143, 1998.
4. Mulroy, S.J, Gronley, JK, Newsam, CJ, and Perry, J: Arch. Phys. Med. Rehabil. 77:187-193, 1996.
5. Boninger, ML, Waters, RL, Chase, T, et. al. Paralyzed Veterans Am. Consortium Spinal Cord Med. 2005.

O-22

TOWARDS A NEW PROTOCOL FOR MOTION ANALYSIS OF THE UPPER EXTREMITIES IN HEMIPLEGIC CEREBRAL PALSY

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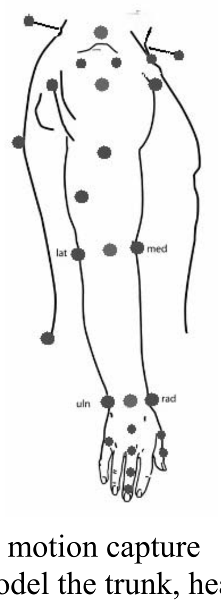
Summary/conclusions
This study describes a model and a testing protocol for motion analysis of the upper extremity in persons with hemiplegic cerebral palsy as a tool in clinical evaluation. A 3D kinematic model was designed and used in motion analysis in 3 subjects with hemiplegic cerebral palsy before and after treatment, and in 3 normal subjects to describe the movement patterns in the upper limb. A reaching task was defined for the subjects to reach, grasp, and retrieve an object placed at various distances on a table directly in front of the subject.

Introduction
The damage to the brain in cerebral palsy (CP) can cause muscle spasticity, involuntary movement, gait disturbance, and abnormal sensation and perception [1]. Spastic cerebral palsy, when occuring in the upper extremities, may affect a person's ability in common reaching and grasping tasks [2]. Spasticity in the upper extremities in children with cerebral palsy can generally upset the balance between antagonist and agonist muscles, wherein, for instance, wrist flexors are often more severely affected than wrist extensors [3]. Treatment often has a goal to restore the balance, and can include occupational therapy, botulinum toxin injections, and orthopedic surgery.

Motion analysis has widespread use in interpretation of gait disorders [4], but no such consensus exists in upper limb models and testing protocols for use in clinical assessment of the consequences of CP on arm function. Furthermore, this tool has yet been used to evaluate the efficacy of treatment in persons with CP.

Statement of clinical significance
A main challenge in upper extremity motion analysis is to define a task that is repeatable, predictable, and functionally realistic, if the motion analysis is to be useful in treatment decision-making or evaluation. If this challenge can be met, motion analysis of the upper extremities may be as clinically useful as gait analysis is to clinicians focusing on the lower extremities.

Methods
Subjects: Three subjects with spastic hemiplegia, (aged 10-13) and three healthy adults participated in this study. Two subjects received injections with botulinum toxin A, both in the biceps and one additionally in the pectoralis, pronator teres, and flexor carpi ulnaris, and were tested 3 weeks after treatment. The third subject with hemiplegia underwent tendon transfer surgery of the hand and arm, including pronator teres rerouting, flexor carpi ulnaris to extensor carpi radialis brevis transfer as well as biceps tendon lengthening, and was tested 2 months after treatment. Passive ROM was measured in elbow flexion and wrist extension.



Model: Subjects performed reaching tasks and were analyzed using a 3-D motion capture system (Vicon Peak). 26 markers were placed over bony landmarks to model the trunk, head, clavicle, humerus, forearm, hand, 2 thumb, and 2 middle finger segments. Euler angle conventions in an x-y-z order were applied to define joint angles. One degree of freedom was assumed at the finger joints, 2 at the wrist (flex/extension and radial/ulnar deviation), 2 at the elbow (flexion/extension and wrist pronation/supination), and 3 at the shoulder (flexion/extension, abd/adduction, rotation). Shoulder girdle, trunk, and head movements were also calculated with 3 degrees of freedom.

Task: Each subject began with the hand on the table, near the navel, and reached for an object placed directly in front of them, retrieving it back to the starting point. 26 trials were performed with each limb, with object distance randomly varied. 10 trials were performed with object distance at 100% arm length from the torso, and 2 trials were performed at each other 10% increments from 50% to 130% arm length. Subjects were instructed to limit trunk use if possible. Additionally, subjects were asked to perform a single trial for each limb to explore their overall range of motion by moving their arm as far as possible in all directions.

Results

The 10 trials performed at 100% arm length from the healthy subjects provided a control curve for each motion in each plane. The subjects with hemiplegia tended to have more elbow flexion, wrist pronation and ulnar deviation. Treatment effects were apparent in many cases. The motion analyses at the different reaching lengths showed that subjects compensated with the trunk and maintained elbow flexion greater than that measured in the passive ROM tests, even when the object was relatively close. Volumetric ROM was lower in the subjects' impaired sides.

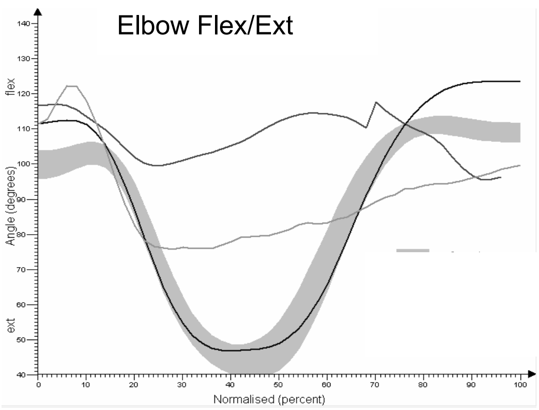


Figure 1: Elbow flex/extension throughout the reach/return task. Control ± 1SD is shown in grey, and the three subjects' motions are shown as individual lines.

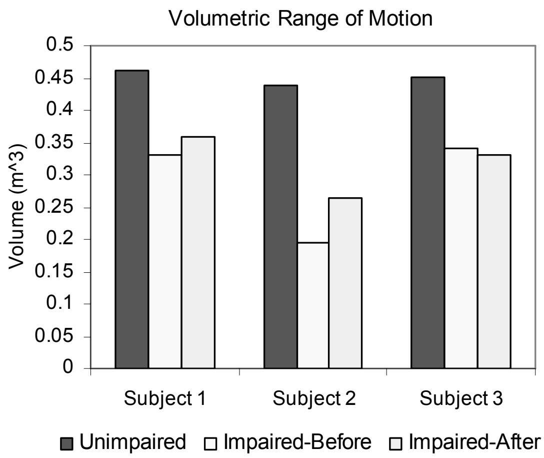


Figure 2: Entire reachable volume in the 3 subjects with hemiplegia.

Discussion

Motion analysis has potential use as a clinical evaluation tool for treatments in the upper extremities, but it is crucial that the functional tasks chosen to analyze (in this case, reaching straight forward and an entire range of motion test) are appropriate. This study is part of a larger attempt to apply concepts from gait & motion analysis to upper extremity treatment.

References

[1] Cooper et al. (1995). J Child Neurol 10:300-309.
[2] Albright (1996). J Child Neurol 11:S1-S4.
[3] Pontén (2005). Dev Med Child Neurol 47(6):384-389.
[4] Gage (2004) Clin Dev Med No 164-165.

O-23

KINEMATIC ANALYSIS OF UPPER LIMBS IN CEREBRAL PALSY SUBJECTS

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Summary/conclusions
The aim of this study was to define a protocol for the quantification of upper limb movements in normal subjects and Cerebral Palsy (CP) patients. The defined protocol in terms of marker positions, upper limb movements, and significant indexes, permitted to characterize the upper limb movements of patients with CP. Furthermore its simplicity let it be suitable for clinical application.

Introduction
CP is characterized by a disruption of motor skills, with symptoms such as spasticity, paralysis, or seizures. Disorders paired with CP include disorders of walking, hearing, eyesight, epilepsy, perception of obstacles, speech difficulties, and eating and drinking difficulties [1]. Functional evaluation of subject movement disorders is very useful in order to define subject status and treatment results [2]. The 3D multifactorial and integrated analysis of movement allows to define quantitatively the motion pattern of a subject in a non-invasive way. This kind of analysis is well known in gait [2], while it is not well defined in upper limb movements, that are not repeatable or cyclic as gait [3].

Statement of clinical significance
The analysis of upper limb movements is useful for the evaluation of pathologies that have as consequence a limited functional movement or kinematical abnormalities in upper limbs, such as CP. Not only this analysis is applicable on non walking subjects – i.e. can be applied when gait analysis cannot - but also it takes in account daily life movements thus resulting very interesting in upper limb functional evaluation.

Methods
The study participants consisted of 14 normal subjects with no history of musculoskeletal or neurological problems, representing a control group (6 males, 8 females; age: 23.2 ± 3.0 years) and 13 patients (6 males, 7 females; age: 14.8 ± 6.2 years) with a diagnosis of CP (6 hemiplegics, 4 tetraplegics, and 2 diplegics). Upper limb movement analysis was conducted using a 12-camera optoelectronic system with passive markers (ELITE2002, BTS, Milan, Italy) working at a sampling rate of 100 Hz, to measure the kinematics of movement [4] and a synchronic Video system (BTS, Milan, Italy). 21 retroreflective markers were placed on each subject's head, trunk and upper limbs. A total of 8 body segments were defined: trunk, right and left upper arm, right and left forearm, right and left hand, and head. Two upper-limb functional tasks were examined: frontal reaching movement and lateral reaching movement. Each subject was seated on an adjustable stool with a table positioned in front of him/her with elbow flexed to 90°. The target was a button to push at a distance. This distance was measured individually for each subject as the 80% of subject arm length for frontal movement and 90% of subject arm length for lateral movement. Each movement was repeated 3 times for each arm, thus each subject completed 12 trials (6 frontal and 6 lateral, right and left).
Trajectory and velocity data of shoulders and hands were studied. We computed index of trajectory deviation from a straight line (IC), as the ratio of the actual three-dimensional length of the path travelled by the endpoint (L_{tr}) to the length of the straight line joining the initial and final endpoint positions (L_0): $IC = L_{tr} / L_0$. Other considered values were movement segmentation, finger covered distance, angles between upper limb segments and their range of

motion. Statistical analysis was conducted using parametric and non parametric tests ($p<0.05$).

Results

In the following only results about IC were reported since it was able to characterize upper limb movements providing clinically significant results (Table 1). In particular IC in control group subjects suggested they had quite linear trajectory (IC frontal movement: 1.35 ± 0.12 ; IC lateral movement: 1.29 ± 0.14) as a matter of fact the more linear the trajectory is the more IC tend to 1. Diplegics were not statistically different from control group subjects (IC frontal movement: 1.5 ± 0.29; IC lateral movement: 1.25 ± 0.14) while the pathological limb of hemiplegics (IC frontal movement: 2.42 ± 0.58; IC lateral movement: 1.86 ± 0.36) and both limbs of tetraplegics (IC frontal movement: 2.16 ± 0.45 ; IC lateral movement: 2.22 ± 0.15) were far away from linearity. Hemiplegic healthy limb had trajectory statistically different from control group (IC frontal movement: 1.65 ± 0.17 ; IC lateral movement: 1.47 ± 0.16) but also more linear than pathological limb. All groups but the tetraplegics presented frontal IC greater than lateral IC.

	IC Frontal Movement		IC Lateral Movement	
	Mean	SD	Mean	SD
Control Group	1.35	0.12	1.29	0.14
Hemi-healthy limb	1.65*	0.17	1.47*	0.16
Hemi-pat limb	2.42*	0.58	1.86*	0.36
Tetraplegics	2.16*	0.45	2.22*	0.15
Diplegics	1.50	0.29	1.25	0.14

Table 1. Mean values (MEAN) and standard deviation (SD) of index of trajectory deviation from a straight line (IC) for both frontal and lateral movements (* $p<0.05$).

Discussion

Choosing IC as characterizing index, diplegics resulted not different from control group, as a matter of fact the upper limb extremities are not involved in this pathology. Tetraplegics resulted the most compromised group since they presented higher IC mean values for both movements than control group. Both arms of hemiplegics resulted dissimilar thus indicating also less compromised limb was interested by movement disorders. This agrees with the fact that hemiplegia shows asymmetrical disorder and the healthy limb usually adopts compensation strategies. Differences in lateral and frontal indexes (frontal IC was greater than lateral IC) in all groups but the tetraplegics, suggested that frontal movement was more difficult than lateral one.

References

1 Murphy N, Such-Neibar T. Cerebral Palsy diagnosis and management: the state of the art. Current Problems In Pediatric And Adolescent Health Care 2003;33:146-169.

2 Gage JR. The treatment of gait problems in Cerebral Palsy. 2003, London: Mc Keith Press; p 1-41.

3 Rau G, Disselhorst-Klug C, Schmidt R. Movement biomechanics goes upwards: from the leg to the arm. Journal Of Biomechanics 2000;33:1207-1216.

4 Ferrigno G, Pedotti A. Elite: a digital dedicated hardware system for movements analysis via real time TV-signal processing. IEEE Trans Biomedical Engineering 1989; 11: 943-50.

O-24

KINEMATIC UPPER LIMB ANALYSIS IN STROKE PATIENTS UNDERGOING CONSTRAINT-INDUCED MOVEMENT THERAPY: 3-MONTH FOLLOW-UP

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Summary/conclusions

The feasibility of our kinematic upper limb evaluation protocol on monitoring the effects of CIMT was tested on 6 chronic stroke patients. CIMT has shown to enhance motor function. Our data confirm motor enhancement up to 3-month follow-up. They also indicate that motor functional recovery is associated with improvement of the selective motor control. This may be a first step for understanding the mechanism underlying improvements due to CIMT. More studies are needed to clarify mechanism of CIMT-induced improvements.

Introduction

A large part of stroke patients are limited in the performance of activities of daily living due to motor impairment of the affected upper limb. One approach to improve motor performance is Constraint-Induced Movement Therapy (CIMT) [1], a rehabilitation regime that has already been shown to enhance motor function in chronic hemiparetic stroke patients [2]. The underlying mechanism(s) responsible for improved motor function of patients undergoing CIMT is still not clear. A possible reason for this lack of understanding may lie in the methods used in assessing upper extremity functions. Most studies rely on clinical tests as Wolf Motor Functional Test (WMFT), Fugl-Meyer, Motor Activity Log (MAL) while more objective outcome measures should be necessary to understand the mechanisms underlying stroke motor deficits and CIMT [3]. Hence, the purpose of this study is to verify if kinematic upper limb analysis may be a feasible method for studying the intrinsic processes underlying motor function improvement in stroke patients after CIMT and for monitoring the results of the therapy in the follow-up.

Statement of Clinical Significance

The proposed kinematic evaluation protocol provides objective measures of upper limb daily living movements like bringing the hand to the mouth [4] and reaching for an object [5]. Particularly, it allows us to quantify the degree of coordination through the computation of normalized jerk [6]. These measures may be helpful to better characterize the functional recovery following CIMT and to gain insight in the mechanisms leading to the improvement.

Methods

Six chronic hemiparetic stroke patients (3 left, 3 right, 4 female, 2 males, average age 49±4 years, 21±5 months since stroke) were selected for this study following the criteria described in Winstein et al. [7]. The group underwent two weeks of CIMT in which the unaffected hand was placed in a mitt. Patients were clinically and instrumentally evaluated before treatment, after 2 weeks of CIMT and again 3 months after.

Clinical evaluation: WMFT (score and time of execution) [8], (MAL) Amount Of Use (AOU) and Quality Of Use (QOU) [1]

Instrumental evaluation: kinematic upper limb analysis during consecutive cyclic Hand To Mouth (HTM) and Reaching (RCH) movements. Instrumentation, kinematic model and acquisition protocol are described elsewhere. [4][5].

Results

In Table 1 clinical and main kinematic outcome measures (average and 1 standard deviation) are shown. Particularly, Movement Duration (MD), Angle at Elbow at end movement (AE), mean Angular Velocity at Elbow (AVE), Coefficient of Periodicity of the Acceleration (CPA)

[9] and Normalized Jerk (NJ) are reported both for HTM (index 1) and RCH movements (index 2). Maximum Angle of Arm Flexion (AAF) is also shown.

Table 1 Clinical and Instrumental Results

		pre CIMT	post CIMT	3 months after CIMT	Unaffected limb
CLINICAL EVAL.	MAL aou	40±23	87±26 *	77±28 †	
	MAL qou	46±22	82±21 *	75±27 †	
	WMFT	57±7	59±8	58±6	
	WMFT-time	2.7±1.0	1.8±.5 *	2.0±.5 †	
HAND TO MOUTH	MD_1 [s]	1.27 ± .14	1.07± .15 *	1.03 ±.15 †	.96 ±.20 §
	AE_1 [°]	131 ± 4	130 ±4	131 ± 6	137 ±4 §¶#
	AVE_1 [°/s]	36 ± 8	41 ± 8	42 ± 7	58 ± 21 § #
	CPA_1 [-]	.80 ± .07	.84 ±.10	.87 ± .07	.91 ±.04 §
	NJ_1 [-]	35 ± 10	24 ±10	23 ± 9	23 ± 7 §
REACHING	MD_2 [s]	1.61 ± .37	1.27 ±.31 *	1.31±.24 †	1.10±.10 § #
	AAF_2 [°]	76 ± 11	74 ± 9	82 ± 8 †‡	81 ± 3
	AE_2 [°]	37 ± 6	42 ± 15	40 ±10	23 ± 5 §¶#
	AVE_2 [°/s]	29 ± 9	36 ± 17	37±12 †	58±5 §¶#
	CPA_2 [-]	.81 ± .10	.89 ± .05 *	.92 ± .02 †	.95 ± .02 §¶
	NJ_2 [-]	63 ± 36	34 ± 17 *	35 ± 10 †	30 ±2 §

Note: symbols indicate $p<0.05$ at the Wilcoxon test:
* pre CIMT vs. post CIMT, † pre CIMT vs. 3 months after CIMT, ‡ post CIMT vs. 3 months after CIMT,
§ Unaff limb vs. pre CIMT, ¶ Unaff limb vs. post CIMT, # Unaff limb vs. 3 months after CIMT

Discussion

Immediately after CIMT all patients show an improvement in MAL AOU, MAL QOU and WMFT-Time scores. Even though some patients show a little regression three months since CIMT with respect to the results right after the therapy, scores are still higher than before treatment. These data are in agreement with Schaechter et al. [2] who found that CIMT-induced motor recovery persists at 6-month follow-up, even if our patients show high functional level, as evident from WMFT scores. Instrumental results, right after CIMT and 3 months later, display evidence of an improvement in all variables but that related to the range of movement. Particularly, 3 months after CIMT, no difference in CPA1,2 and NJ1,2 is discernible anymore among paretic and unaffected limbs. Since CPA is a measure of the repeatability of the movement [9] and NJ a measure of degree of coordination [6], this may be an indication that clinical evidence of motor recovery due to CIMT is related to improvement of the patients' selective motor control. More patients should be studied to verify the correctness of this supposition. Main conclusions refer to the feasibility of our method to monitor the effect of CIMT in the follow-up.

References

[1] Taub E et al., (1993), Arch Phys Med Rehabil., 74:347-54
[2] Schaechter JD et al., (2002), Neurorehabil Neural Repair., 16(4):326-38.
[3] Alberts et al., (2004), Neurorehabil Neural Repair., 18(4):250-8.
[4] Molteni, et al.,(2003) Eur Med Phys.,39(suppl 1 to 3):89-91.
[5] Molteni, et al., (2004) Eur Med Phys., 40(suppl 1 to 3):320-2.
[6] Teulings H.L. et al.,(1997), Exp.Neurology, 146: (159-170),
[7] Winstein CJ et al.,(2003) Neurorehabil Neural Repair, 17(3):137-52
[8] Morris DM et al., (2001), Arch Phys Med Rehabil., 82:750-755
[9] Sabatini AM., (2002), Biol Cybern., 86(4):253-62

O-25

3D KINEMATICS OF UPPER EXTREMITY TASKS

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Summary/conclusions
This study presents a measurement method for the 3D kinematics of the Upper Extremity (UX). Also a series of standardised UX tasks and their kinematic ranges are presented. These results are the basis for a 'UX analysis report' compared to the 'gait analysis report'.

Introduction
Though the kinematics of the UX have been studied before [1], these studies did not include sets of standard functional movements of the UX that included the correct hand orientation. Since hand orientation is a major determinant in the usefulness of upper limb motion, this is a serious drawback. The purpose of this study was firstly to define a measurement method based on current state of the art that includes hand orientation. Secondly, to define a set of functional tasks for the UX and establish the stereotype execution of the tasks (norm values) as well the amount of normal variation.

Statement of clinical significance
The development of gait analysis has been extremely useful in the treatment of lower extremity dysfunctions. Likewise, analysis of UX functions by means of 3D kinematics has the potential to become an important tool in clinical decision making and therapeutic evaluation of patients with UX disorders.

Methods
Movements of the UX were measured using stereophotogrammatic recording of active LED-markers using an Optotrak (Northern Digital) system. Small clusters of 3 markers were placed on the hand, upper arm, acromion (to represent the scapula [2]) and trunk. Also, a cuff with 6 markers was placed just proximal of the styloids of the wrist. Local anatomical coordinate systems and joint rotations were defined according to the ISB standardization proposal [3]. The proximal landmark for the humerus was determined using a helical axis approximation. The orientation of the humerus was defined with respect to the thorax. Axial rotation of the humerus was determined using the orientation of the forearm. Zero angles were defined by alignment of the anatomical coordinate systems.
Ten healthy adults (age 23-42) performed 6 simple whole ROM tasks (wrist palmar - dorsal flexion, pronation - supination, elbow flexion - extension, endo - exorotation with 90^o humerus abduction, anteflexion - retroflexion, abduction - adduction). Additionally they performed four functional tasks (drinking, combing hair, move hand to back pocket, move hand to contra lateral shoulder). All tasks were performed three times and recorded at a sample rate of 50 Hz. For the ROM tasks the subjects started in the anatomical position and were asked actively move to a maximum joint angle in that plane. The four functional tasks were selected after consultation with the clinical staff. To promote standard performance, subjects were asked to copy the movements made by the instructor who stood in front of them. From the ROM tasks the following maximum angles were selected for further analysis: wrist palmar and dorsal flexion, pronation, elbow flexion, humeral endo- and exorotation, humeral elevation and scapula laterorotation. The same angles were derived for the functional tasks, from the starting

point to the endpoint of the movement and time was normalized (0-100%).

Results

The results from the ROM tasks are given in figure 1. Maximal humeral elevation during anteflexion (138° ± 9°) and abduction (133° ± 9°) are very close. Maximal scapula laterorotation during anteflexion also appeared to be close to values during abduction (51° ± 12° versus 55° ± 14°). For the four functional movements a large set of kinematic data was obtained, as an example the task drinking is chosen to be presented here. Figure 2 shows the joint angles ± SD during the movement from the starting point (hand on the lap) to the endpoint (drink from the cup) of the task.

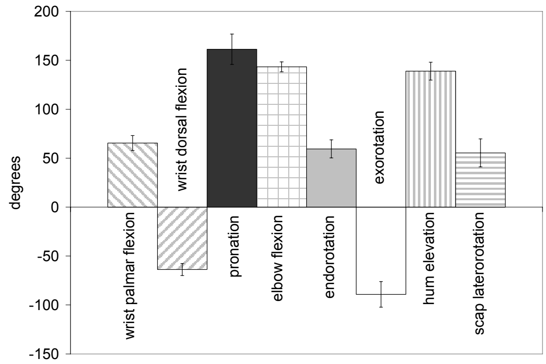

Figure 1. Average values of ROM ± SD.

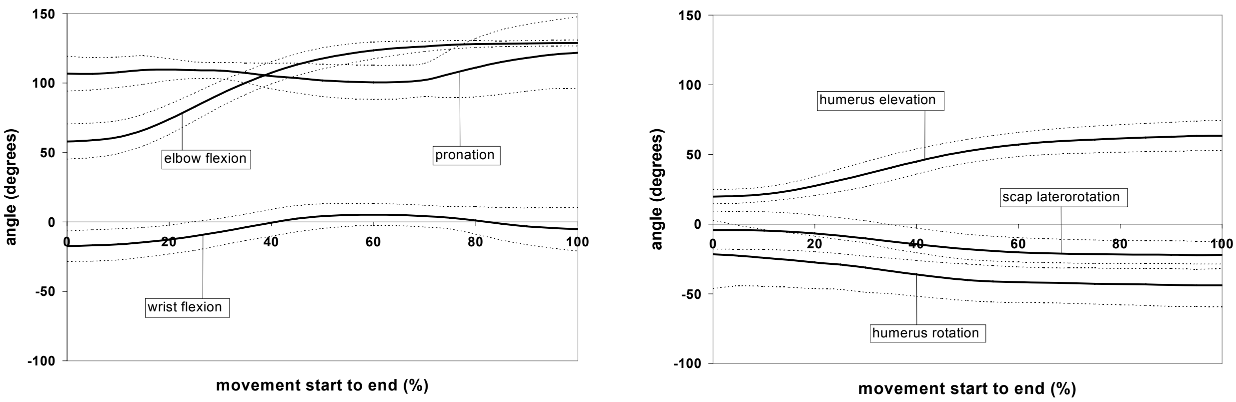

Figure 2. Average joint angles ± SD during the functional task drinking.

Discussion

The developed functional evaluation method was technically feasible for healthy adults. ROMs of shoulder and elbow were consistent with earlier results [4]. The joint angles during the functional task show that the subjects use similar trajectories to reach the end goal of the task, i.e. the norm values of the task can be determined. Humeral endorotation, however, shows a larger variation during the whole trajectory. The variation in pronation increases near the end of the task, possibly due to differences in cup filling. Further study will reveal whether the method can discriminate between healthy subjects and patients and if compensatory movements can be detected. In conclusion the method defines a basis for a clinical 'UX analysis report'.

References

[1] Anglin C. and Wyss U.P., 2000, Proc Instn Mech Engrs 214 Part H: 541-55
[2] Karduna A.R. et al, 2001, J Biomech Eng 123(2): 184-90
[3] Wu G. et al., 2005, J Biomech 38: 981-92
[4] Magermans D.J. et al., 2005, Clin Biomech 20(6): 591-9

O-26

CLINICALLY SIGNIFICANT SHOULDER JOINT KINEMATICS DESCRIPTION

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Summary/conclusions
While there are numerous methods available for the calculation of shoulder joint kinematics, the ability to present these in clinically significant terms has remained a challenge. Through realignment of coordinate axes on the humerus, shoulder kinematics can be calculated in terms that are familiar in clinical settings.

Introduction

The ability to describe calculated shoulder joint kinematics in clinically relevant terms has, to date, been difficult. This difficulty has likely been compounded by the plethora of methods used to decompose the rotation of the shoulder joint. Rotations captured using motion analysis techniques are analysed most widely using either Euler angle or screw axis decomposition. While those who use Euler angles in their analyses may choose to follow the rotation sequence initially proposed by An [1] and recommended by the ISB [2], This method, however, offers shoulder motion results in terms of the angle and amount of humeral elevation in lieu of outright measures of flexion/extension or ab/adduction. Additionally, in the clinical setting, neural, muscular, or skeletal disorders may cause individuals to take up postures and compensations that may make recommended coordinate assignments or decompositions provide unwanted results. Our laboratory has previously advocated the use of screw-axes for the analysis of shoulder motion to avoid some of the pitfalls of gimbal lock that may be associated with Euler angle decomposition [3]. However, the problem still remains that shoulder flexion/extension and ab/adduction are measures of the humerus about axes of the trunk. Internal/external rotation of the humerus can change the alignment of the humeral sagittal and coronal axes, confounding the ability to examine the motion of the humerus with respect to the trunk (or scapula, for glenohumeral motion). The current study outlines a method for presenting shoulder kinematics in terms of clinical rotations.

Statement of clinical significance

A simple correction algorithm may be applied to the decomposition of shoulder joint kinematics to allow for clinically meaningful interpretation of shoulder joint motions.

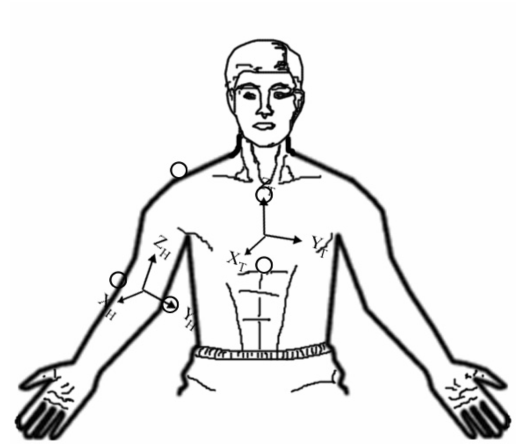

***Figure 1**. Schematic showing marker locations and coordinate axes for humerus and trunk coordinates systems. Not shown is the C7 marker.*

Methods

This study was conducted as part of a larger study for the creation of a database of normal kinematics during a set of motions used in clinical upper extremity assessments of the neck, scapulothoracic, glenohumeral, elbow, and wrist joints, as well as the thoracohumeral pseudo-joint. Subjects were asked to perform select motions (full shoulder flexion, abduction, touching their head, touching their opposite shoulder, etc.) in a comfortable motion. For the purposes of this discussion, only motion of the humerus with respect to the thorax during shoulder flexion will be considered, due to space constraints. The motion of retro-reflective markers was tracked at 60 Hz using a 10-camera Motion Analysis EVa RealTime system (Motion Analysis Corp., Santa Clara, CA, USA). Three markers, on the xiphoid process, sternal notch, and C7, were used to define the trunk

(Fig. 1). Three markers on the acromion and medial and lateral epicondyles tracked motion of the humerus. Coordinates for these segments were formed consistent with ISB recommendations [2]. A static (anatomic) pose is captured with the elbow fully extended, the arm at neutral abduction and flexion, and the palm facing the subject's thigh. The local (humeral) coordinate system (HCS) is transformed into the clinical flexion/extension-ab/adduction coordinate system (CCS) for both the static and dynamic (activity) trials. The CCS is comprised of the previously defined humeral Z_H, and transformed x'- and y'-axes that are defined as follows: (1) the x'-axis is taken to be the intersection of the humeral X_HY_H plane and the thoracic X_TZ_T plane, which is, by definition, perpendicular to the humeral z-axis; (2) the y'-axis is the cross product of the humeral z- and x'- axes. Relative rotations are calculated between the Thoracic and CCS, and helical-axis decomposition is used to decompose the rotation matrices as described previously [3]. The components of the helical axis in the CCS x'-, y'-, and z-axes will represent, respectively, the amount of clinical abduction, flexion, and internal rotation for a given shoulder motion.

Results

Results of an able-bodied subject performing five shoulder flexion/extension cycles are shown (Fig. 2). Decomposition of the angles in the HCS revealed 75 degrees of peak rotation about the HCS X_H-axis and 115 degrees of peak rotation about the HCS Y_H-axis. Alternatively, the measures of maximal abduction were found to be around 42 degrees, while there was a maximum 144 degrees of shoulder flexion. The second approach more closely represents the motion observed clinically. While there was still some amount of ab/adduction, this is due to the fact that the subject was not constrained in their chosen path during flexion/extension trials.

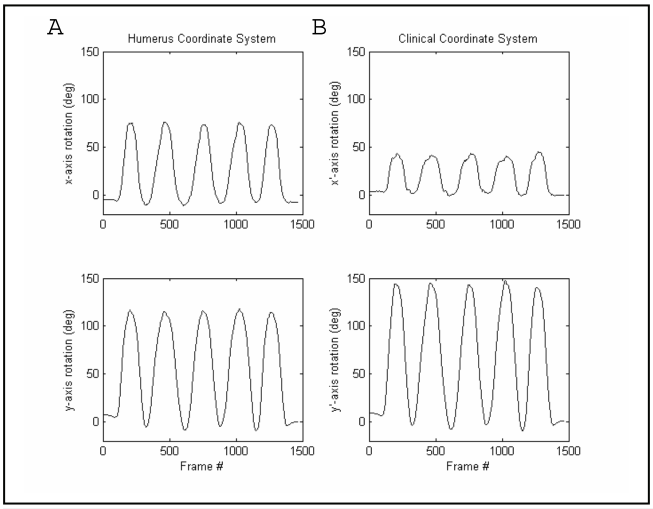

Figure 2. *Kinematics from a subject performing shoulder flexion trials for helical-axis decomposition of the motions in (A) the HCS, and (B) the CCS.*

Discussion

While use of decomposition of the relative rotations of shoulder motion generated using the HCS yields correct and useful information for the purposes of generating biomechanical models, this data does not represent the motion in terms that are always meaningful in the clinical setting. Use of the proposed CCS for decomposing shoulder motion transforms relative motions from those occurring about humerus-fixed axes to axes that relate to the thorax are more clinically relevant. Additionally, since this method uses helical-axis decomposition, there is no need to alter rotation orders to compensate for the possibility of gimbal-lock due to either prescribed motions or unanticipated compensations. While this method is presented using relative motion of the humerus with respect to the trunk, it can easily be applied to examination of glenohumeral motion as well.

References

[1] An, Kai-Nan, (1991), Journal of Orthopaedic Research, 9, 143-9.
[2] Wu, GE, (2005), Journal of Biomechanics, 38, 981-992.
[3] Morrow, DA, (2003), Gait & Posture, 18, S35-S36.

O-27

OBJECTIVE IDENTIFCATION OF GAIT ABNORMALITIES

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Summary/conclusions
Objective identification of kinematic gait abnormalities was implemented on six consecutive patients at the Oxford Gait Laboratory. This was found to be a useful method for improving consistency of interpreting gait reports and has potential to enhance reliability of treatment recommendations.

Introduction
The repeatability of treatment recommendations based on clinical gait analysis has recently been questioned [1]. An element of this variability may be attributed to lack of consistency in identifying gait abnormalities. A method to automatically generate a list of gait deviations is proposed here and implemented in six individual case studies.

Statement of clinical significance
Automatic identfication of gait abnormaliteis aids repeatability of interpreting gait reports and has potential to produce more consistent treatment recommendations.

Methods
Five sets of kinematic graphs obtained from clinical patients following three-dimensional gait analysis (Vicon) were reviewed during a routine interpretation session by a consultant orthopaedic surgeon, a physiotherapist, a bioengineer, an orthotist and a clinical technologist. Visual identification of abnormality was performed through a systematic process and results were recorded in a table. The same five sets of graphs were then submitted to an automated assessment of abnormality, using the Parameter Check PlugIn (Vaquita Software). An initial list of kinematic variables was chosen for analysis and compared to the visual assessment of kinematic graphs. This list was further refined based on initial results to reflect clinical relevance. Mean and standard deviation values were determined from the kinematic graphs of 36 healthy children. If the difference between recorded value and the healthy mean was greater than one standard deviation, then the variable was deemed to be abnormal and recorded as such. The percentage agreement between the visual observation and automatic list of abnormalities was calculated. In addition, the kinematic graphs obtained from one patient were visually reviewed on two separate occasions, to determine percentage of agreement between visual identification of abnormality on different days.

Results
An example of kinematic graphs from a child (age 10 years, 11 months) with spastic diplegic cerebral palsy is shown in figure 1. The abnormalities which were identified automatically were ordered according to the magnitude of the difference from normal and are listed below:

1. Increased knee flexion at initial contact (no. of SDs)
2. Reduced maximum knee extension in stance
3. Reduced maximum hip extension
4. Exaggerated hip flexion in swing
5. Increased average foot external progression
6. Exaggerated peak knee flexion in swing
7. Pelvic obliquity with the right side down
8. Reduced range of dorsiflexion

9. Reduced average dorsiflexion in stance
10. Increased average foot adduction

Comparison of visual and automatic identification of kinematic gait abnormalities revealed an average of 74% agreement across the six case studies.

Inter-day repeatability of visual assessment of gait patterns was also determined and revealed comparable agreement of 76%, when selecting abnormalities from a pre-defined list.

Discussion

Automatic identification of abnormalities produced more consistent results than visual observation alone. This was particularly apparent for variables represented by averages and ranges, rather than maximums and minimums. Differences less than 1.5 standard deviations away from normal were also more consistently detected by the automated system than by visual observation. However, overall abnormal kinematic patterns were not always reflected in the specific variable studied. Improvement in consistency of interpretation was observed, however reviewing of the kinematic graphs is still necessary, in combination with the automated output, to determine which deviations are clinically relevant and to explain the significance of each of the findings. For example, both positive and negative differences between the healthy mean and case study results were detected with the automated system. However, it is frequently the case that only change in one direction is meaningful and suggests the necessity of intervention. Care is also required to ensure that other abnormalities, which are not included in the automatic list, are also identified. Work is ongoing to include kinetic and EMG variables in this analysis, and to produce a finite list of potential gait deviations on which to base consistent interpretation of results.

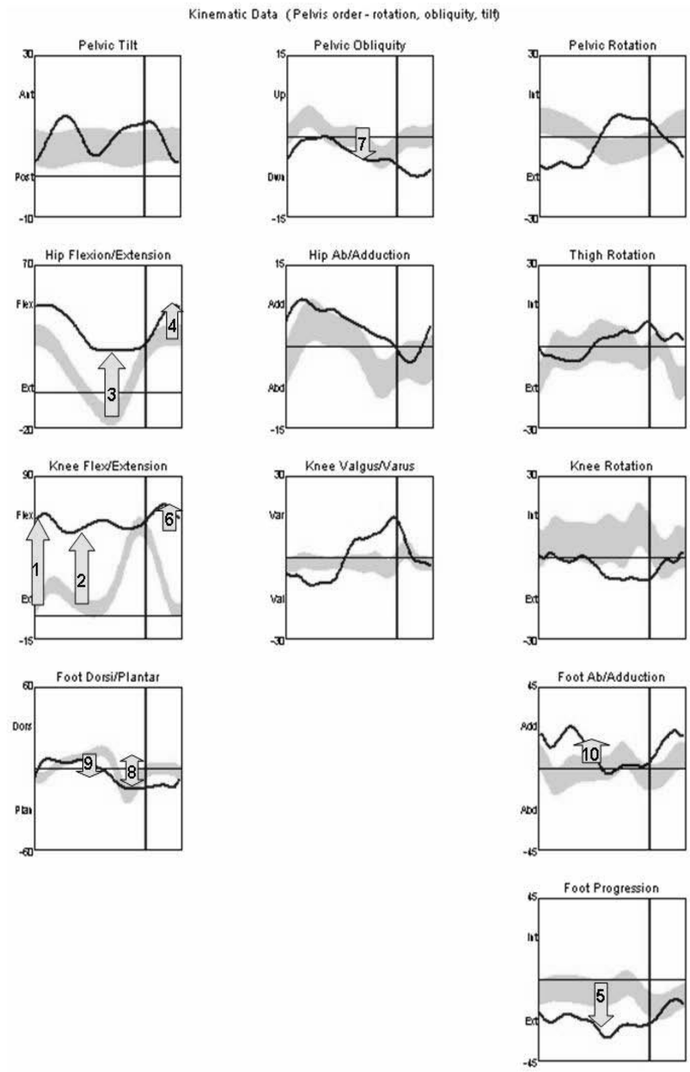

Figure 1. *Example kinematic graphs from the right leg (blue line) of a child with CP (diplegia). The green band represents mean graphs from the healthy children +/- 1SD*

References

[1] Noonan, K.J., et al., (2003), J Pediatr Orthop, 23, 279-87

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NORMALCY GAIT INDEX AND KINEMATICS: UNCERTAINTY AND REPEATABILITY ON HEALTHY CHILDREN DATABASE. PRELIMINARY APPLICATION ON CEREBRAL PALSY GROUP.

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Summary/conclusions
A large database of asymptomatic gait children was established. Repeatability and uncertainty of normalcy gait index and kinematics using Helen Hayes protocol were evaluated. This study should allow more relevant comparison between patients and healthy children, and between pre and post treatments.

Introduction
Most of Cerebral Palsy studies use Helen Hayes protocol [1] to calculate kinematics parameters. Recently, Normalcy Gait Index [2] was proposed to estimate deviation of a subject's gait from a normal profile. This index shows potential as a useful tool to objectively quantify overall changes in patient's gait.

Statement of clinical significance
Repeatability of normalcy gait index has not been evaluated yet, and repeatability of Helen Hayes protocol was primarily evaluated for adults [3]. As the technique and associated parameters become the basis of gait quantitative evaluation, it is important to estimate their uncertainty to help in clinical interpretation. The aim of this study is to evaluate uncertainty and repeatability of kinematics parameters and normalcy gait index, and to constitute a database of healthy children. A preliminary clinical application was performed on a small group of cerebral palsy children.

Methods
56 asymptomatic children (28 boys, 28 girls) aged between 5 and 15 years old (mean: 10) have performed the gait exam (VICON® system MX3) using the Helen Hayes protocol. 29 kinematic parameters [4] were extracted from lower limb joint angles curves. Normalcy gait index (NI), using the same parameters as Shutte et al. [2], was also calculated on this database. 16 subjects performed the exam twice with markers replacement to assess repeatability of parameters. Uncertainty was assessed as 2SD of differences to mean values between the 2 sessions. A preliminary clinical application was performed on 8 cerebral palsy children (mean age: 9 years old). 3 patients were quadriplegics, and 5 were diplegics. Different trials were collected when patient was used to use one of these assistances during his walk: key-walker, orthosis, canes and shoes. A simple gait is acquired when patient can walk without technical assistance. As trials could not be repeated for patients, Monte Carlo simulations were carried out by adding a noise measurement on each kinematic curve used in NI computation, to evaluate its uncertainty. These simulations were done on asymptomatic and cerebral palsy subjects. 2SD values of uncertainty are presented.

Results
A database of healthy subjects with "corridors" of normality was established for kinematics and kinetics curves. NI calculated within asymptomatic children ranged between 5 and 30, with

a mean value of 15.4 (right and left indices averaged). Uncertainty evaluated by repeatability studies within 16 healthy subjects was ±2° for hip angles in sagittal plane (95% of confidence interval), ±3° in the frontal plane and ±6° in the horizontal plane. Uncertainty for Knee and Ankle was ±5° (in sagittal plane). Time of events had a repeatability of 4%. Repeatability of NI was approved (p=0.071) with ±6 as confidence interval at 95%. Joint angles and NI were calculated within the 8 cerebral palsy children and compared to our database. Results of Monte Carlo simulations are given in Table 1. Uncertainty (2SD) on NI was linearly correlated with NI values (R^2 = 0.883), and could be estimated for patients using the relation $\Delta_{NI} = 14.877 + 0.043*NI$.

Table 1. Monte Carlo Simulations on Normalcy Gait Index: healthy children and Cerebral Palsy patients

		Type of walk	*Normalcy Index*	*2SD (Monte Carlo simulations)*
Asymptomatics	56 Subjects	normal	15.4 (5-30)	6
Diplegics	Patient 1	normal	38	9
	Patient 2	normal	296	30
	Patient 3	key-walker	670	40
		canes	795	53
	Patient 4	key-walker	870	34
	Patient 5	normal	800	56
		orthosis	603	52
Quadriplegics	Patient 6	normal	389	39
		key-walker	385	34
	Patient 7	normal	1137	65
		orthosis	496	42
		shoes	482	36
	Patient 8	normal	2000	98

Discussion

Mean value and range of NI were similar to those calculated by Shutte et al. [2] on 24 asymptomatic children. Repeatability found on kinematics was closed to that found by Schwartz et al. [5]. Intervals of uncertainty for angles and NI contain sources of errors due to intrinsic system's error, marker placement, relative movement between skin and markers, and sensitivity of biomechanical model to these errors. These values should be taken into account during comparison between patient and normal subject, and between pre and post treatment for a patient.

Acknowledgment

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References

[1] Davis RB (1990): A gait analysis data collection and reduction. *Hum Move Sci.* 1990; 10:575-87.
[2] Shutte L M (2000): An index for quantifying deviations from normal gait. *Gait and Posture*. 11: 25-31.
[3] Kadaba M P (1989): Repeatability of kinematic, kinetic, and electromyographic data in normal adult gait. *J.Orth. Res*. 7: 849-60.
[4] Goujon H (2005): A functional evaluation of prosthetic foot kinematics during lower-limb amputee gait. *Prosthet. Orthot. Int*. In Press
[5] Schwartz M H (2004): Measurement and management of errors in quantitative gait data. *Gait and Posture*. 20(2): 196-203.

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GILLETTE GAIT INDEX IS CONSISTENT WITH QUALITATIVE VISUAL ASSESSMENTS OF GAIT

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Summary/conclusions
The Gillette Gait Index (GGI) provides quantitative scores that are consistent with qualitative assessments of overall gait for individual patients with cerebral palsy (CP). This supports the validity of GGI as a useful summary measure for gait analysis.

Introduction
Modern gait analysis produces a large volume of data. While these data are highly informative for specific aspects of gait such as velocity or peak dorsiflexion in stance, a quantitative measure describing a patient's overall gait is not usually reported. The Gillette Gait Index (GGI), formerly called the Normalcy Index, has been proposed as a summary measure that takes into account 16 clinically important kinematic and temporal parameters [1]. Previous studies have shown that the GGI can detect differences in groups of subjects who either have different diagnoses or have undergone surgery [2, 3]. However, to fully validate this measure for use in individual patients, studies are needed to evaluate the validity of the GGI on an individual level. The purpose of this study was to compare GGI scores with qualitative assessments of gait in individual patients. This is needed to determine if the GGI provides a reasonable summary score that clinicians and researchers can use as a global measure of gait in individual patients.

Statement of clinical significance
By demonstrating that the GGI is consistent with qualitative assessments of overall gait for individual patients, this study helps to establish validity of this measure for quantifying patient status and clinical outcomes.

Methods
Pre- and 1-year postoperative gait analysis and video data were examined for 25 children with CP who underwent multilevel lower extremity orthopaedic surgery to correct gait problems. The gait analysis data were used to calculate the GGI [1]. 12 raters, including 3 experienced gait laboratory physical therapists (PTs), provided qualitative scores for each subject's gait based on the videotape recordings. While there was no set time period for watching the videos, most raters completed the study in a single session.
First, the raters watched the 25 pre- and 25 post-operative videos mixed together in random order, along with 6 repeats (56 videos total). Each video was rated as 1 = minimal impairment, 2 = moderate impairment, 3 = substantial impairment, 4 = severe impairment. Results are presented for overall walking ability; similar results were obtained for walking efficiency and for stability and balance. Next, the raters watched each patient's pre-operative video followed immediately by the post-operative video. The pre- to post-operative change was rated as 3 = significantly improved, 2 = moderately improved, 1 = slightly improved, 0 = not much better or worse, -1 = slightly worse, -2 = moderately worse, -3 = much worse.
Kappa statistics were used to evaluate intra- and inter-rater reliability of the video ratings. The Kruskal-Wallis test was used to compare the raters' assessments of change when viewing the pre- and post-operative videos separately versus together. Scores between –1 and –3 were combined in the Kruskal-Wallis analysis since only a few scores fell in this range. Linear regression was used to compare the GGI scores with the mean video ratings.

Results

There was moderate agreement between the raters' first and second assessments of the 6 repeated videos (Weighted Kappa = 0.52). Of 72 pairs of ratings, 42 (58%) were exactly the same, 29 (42%) differed by ±1, and only 1 (1%) differed by ±2. Agreement among the 12 raters was fair for assessment of the individual videos (Kappa = 0.25). Agreement was higher among the gait laboratory PTs (Kappa = 0.31) compared with the other raters (Kappa = 0.22) and for scores at the ends of the scale (Kappa 0.32 – 0.52 for scores 1 and 4) than for intermediate scores (Kappa 0.09 – 0.23 for scores 2 and 3). Agreement among the raters was lower for pre- to post-operative change (Kappa = 0.11). Agreement was highest for scores of –1 (Kappa = 0.37) and 3 (Kappa = 0.53) among the gait laboratory PTs although even the PTs had only slight agreement in the intermediate scores (Kappa <0.08).

Raters tended to score patients as having greater improvement when the pre- and post-operative videos were viewed together compared with when the videos were viewed separately. 270 (73%) of the scores given when viewing the videos together indicated improvement, but only 138 (37%) of the pre-operative scores improved post operatively, with 192 (52%) remaining unchanged. Using the mean pre-operative, post-operative, and change scores from all raters, there was a highly significant relationship in the assessment of improvement between viewing the videos separately versus together (P = 0.0001).

The GGI scores reflected the mean video ratings pre-operatively (P = .003), post-operatively (P = .005), and in pre- to post-operative change (P = .006 for absolute change; P = .02 for % change). One subject was a clear outlier and was excluded from the analyses. This subject demonstrated substantial improvements in dorsiflexion which led to a mean score of 1.5 from the raters, but had a worsening in GGI due to deterioration of less visual variables such as range of pelvic tilt and time to peak knee flexion in swing.

Discussion

The raters differed in some individual ratings and tended to note more improvement when viewing the pre- and post-operative videos together. However, using mean scores from the group as a whole, there was good correlation between viewing the videos separately versus together. The mean scores therefore provide a reasonable basis for comparing the GGI scores. The GGI scores were consistent with these mean scores in 24 of 25 patients.

The one outlier demonstrates the difference between the visual and quantitative assessments of gait. While the visual ratings were primarily influenced by the patient's obvious improvement in dorsiflexion, this improvement was outweighed by worsening of several less visual variables in calculation of the GGI. Both assessments are "true"; the difference lies in the relative weight given to different variables. The quantitative assessment is able to account for more variables. Other variables and different weightings could be implemented if desired.

A previous study performed detailed visual analysis of specific gait variables by persons experienced in gait analysis to calculate the Edinburgh Gait Score (EGS) [4]. This study found high correlations between EGS and GGI. Our study differs by assessing gait globally rather than in terms of specific gait variables. Taken together, the two studies suggest that the GGI is a valid measure for assessing overall gait in individual patients.

References

[1] Schutte et al., Gait & Posture 11:25-31, 2000;
[2] Romei et al., Gait & Posture 19:85-90, 2004;
[3] Schwartz et al., J Pediatr Orthop 24:45-53, 2004;
[4] Hillman et al., Proc. 13th ESMAC, p. 124, Warsaw, 2004

Acknowledgements

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GAITABASE: NEW APPROACH TO CLINICAL GAIT ANALYSIS

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Summary/conclusions
A web interfaced repository for gait analysis data, the GAITABASE, has been established. It allows users anywhere in the world to contribute data and to view all the data selected for specific groups of subjects and specific conditions of data capture. It's use has been illustrated in the context of a mock cohort study of children with CP who have undergone single-event multi-level surgery which includes data from 2 centres in USA, one in Europe and one in Australia.

Introduction

Decision-making in clinical gait often relies on comparing gait patterns from populations with and without gait pathology. Such comparison, however, requires dataset from large sample size population differed by age and gender, which in most cases does not exist (1,2,3,4). The objectives of a valid international gait analysis repository will be to allow orthopaedic surgeons, physiotherapists, biomechanical engineers, and human movement specialists to share data from large population, create and select large normative dataset, and to give a valuable insight into how effective specific interventions have been for others. Lately, researches began to build gait analysis repository in their own local laboratories (5, 6). Unfortunately, these repository systems did not provide the capability of sharing gait patterns between national and international clinics and research groups. Moreover, they were still limited by the small patients sample size.

We present here the GAITABASE, a web interface database for gait analysis, that in the future will provide ways of collecting and collating, and accessing data from all participating clinical and research centres around the world. Four main concepts are implemented to ensure that the system is flexible: 1) secure archiving of processed gait data, 2) flexible personalised filter mechanism that allow users to create criteria for querying gait data, 3) visualising the results in appropriate tables and graphs, 4) export query results for statistical analysis.

To illustrate the benefits of GAITABASE we will present a case study of Cerebral Palsy patients where gait analysis was performed before and after an intervention.

Statement of clinical significance

GAITABASE is a valuable resource and tool for the international clinical and research community. It provide ways of sharing data from all participating centres to establish normative and pathology datasets of much larger numbers of patients than could be achieved by any single centre. The accumulated stored data will facilitate clinicians in comparing their own gait data with others, and will give a valuable insight into how effective specific interventions have been for others.

Methods

GAITABASE is web-based interface utilising MySQL database, HTML, CSS, JavaScript, PHP, and Fusebox 3.0, which is hosted by the Royal Children Hospital and accessed through the Murdoch Children's Research Institute web site. Stored data include: 1) patient's data such as code, pathology, gender, 2) temporal-spatial parameters such as walking speed, stride length and cadence (as measured by GaitRite system), and 3) time-normalized 3D kinematic and kinetic data (joint angles, moments, powers) which are uploaded from c3d files. Ethical approval has been obtained to receive, store and publish de-identified gait data. Contributors of

data will have to obtain ethical approval from their own institutions to contribute data.
To illustrate the potential of the GAITABASE and to test whether users in different countries would be able to contribute data, a mock cohort study of progress of children following single event multi-level surgery for children with diplegic cerebral palsy was set up. Users in USA, Europe, Australia and New Zealand were approached to contribute pre-operative and one year follow-up data.

Results

The GAITABASE web interface allowed the user to add, view, edit, import, and export patient's gait data. Users from two centres in the USA, one in Europe and one in Australia contributed data of 5 CP children to the mock cohort study. The Figure below shows pre-operative and follow-up data of one CP child with mean ±1SD of 10 children without pathology.

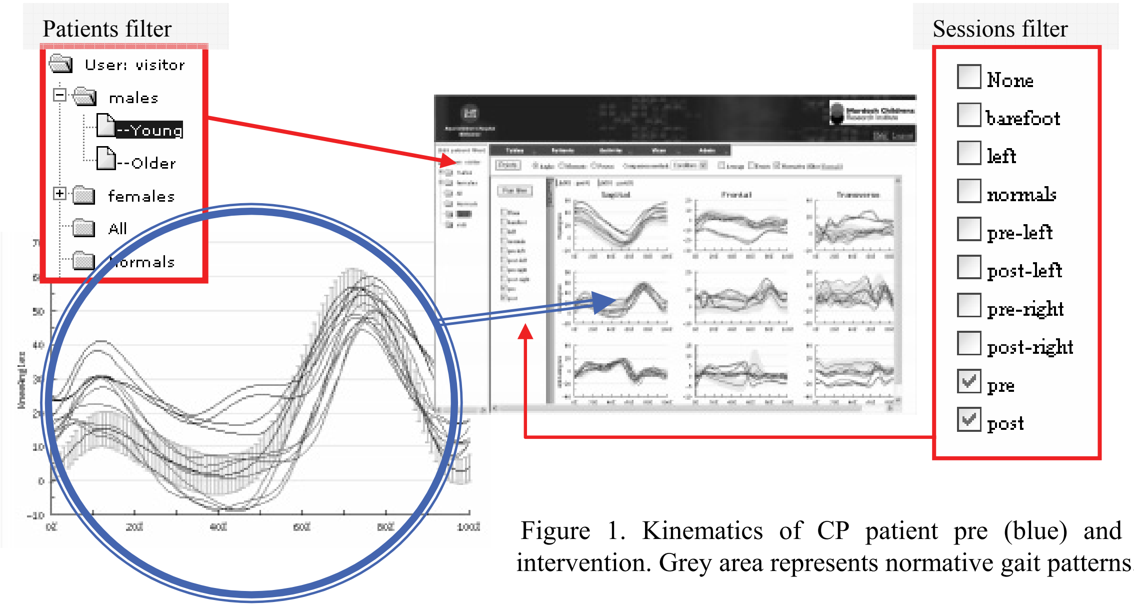

Figure 1. Kinematics of CP patient pre (blue) and post (red) intervention. Grey area represents normative gait patterns.

Discussion

This exercise proved that the concept of the GAITABASE can work. Several of the users approached for gait data envisaged problems in obtaining ethical approval to contribute data but those who set out to obtain this were able to complete the process and release the data within two weeks. Future work on the GAITABASE will include procedures for ensuring the quality of the data stored within it.

References

[1] Winter DA., (1993). Knowledge base for diagnostic gait assessments. Med Prog Technol, 19(2):61-81.
[2] Prince F., Corriveaua H., Héberta R., and Winter DA., (1997). Gait in the elderly. Gait & Posture, 5(2):128-135.
[3] MacWilliams BA., Cowley M., and Nicholson DE., (2003). Foot kinematics and kinetics during adolescent gait. Gait & Posture, 17(3):214-224.
[4] Al-Obaidi S., Wall JC., Al-Yaqoub A., and Al-Ghanim M., (2003). Basic gait parameters: A comparison of reference data for normal subjects 20 to 29 years of age from Kuwait and Scandinavia. J Rehabil Res Dev, 40(4):361-6.
[5] Berghof R. and Zwick E., (1995). Gaitbase - A relational database for the management of gait laboratory data. Gait & Posture, 3(4):291-292.
[6] Schablowski M, Schweidler J, Rupp R., (2004). HeiDATAProVIT-Heidelberg data archiving, tag assembling, processing and visualization tool. Comput Methods Programs Biomed, 73(1):61-70.

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A MODIFIED GAIT CLASSIFICATION SYSTEM FOR CHILDREN WITH HEMIPLEGIA

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Summary/Conclusions

The refinement of a computerised algorithm [1] for the classification of hemiplegic gait by the addition of two criterions: reduced range of dorsiflexion in swing and absence of first rocker, resulted in the classification of 93% (78/84) of individuals. Classification of the walking patterns of an entire population of children with hemiplegia is possible using an automated system.

Introduction

A previous algorithm [1] based on the work of Winters and Gage [2] was unable to classify the gait pattern of 35 out of 84 children with hemiplegia. Visual observation of the unclassified gait patterns revealed minor deviations only, suggesting that the original system lacked the sensitivity to detect mild abnormalities in gait. Thus the need for refinement of the original automated system was manifest.

Statement of clinical significance

Automated detection of common gait abnormalities using objective parameters can be used in the clinical and research setting to assist in the classification of children's walking patterns. This technique affords a labour-saving alternative method to visual observation when evaluating large datasets.

Methods

Participants/Procedure

Valid 3-D kinematic data was obtained for 84 (50 male, 34 female; mean age 10.8 years, age range 5-16 years) out of 94 children with hemiplegia who participated in a population-based study of locomotor ability in cerebral palsy. Of the other 10 children, two refused to co-operate with the assessment procedure and eight had insufficient quality data for processing. Data was captured using a Vicon 612 motion capture datastation and six-camera array. Three gait trials were processed and imported into a Polygon report. A single representative gait cycle was then exported to Microsoft Excel. The original algorithm classified forty-nine children (Table 1), leaving a remaining 35 children as 'unclassified'. The latter group was subject to the following scrutiny and subsequent analysis.

Data Analysis

Visual observation of the 35 'unclassified' gait cycles was carried out by a physiotherapist and research assistant. Two common deviations were noted: (1) reduced range of dorsiflexion (DF) in swing and (2) absence of first rocker. On the basis of this two additional logic statements were written into the original algorithm. For computational purposes these were defined as (1) a range of DF less than 9.87degrees during swing phase, derived from the laboratory's age-matched database of able-bodied children (n=18) and (2) the presence of a negative gradient in the sagittal DF graph over the first four captured frames.

Refined/Modified Classification

The original Type I algorithm (DF<0 during swing) was refined to include the first logic statement (reduced range of DF in swing) and for the purposes of this analysis will be referred to as Type Ib. The classification was also modified to include the second logic statement (absence of 1st rocker), referred to as Type Ia. All 84 datasets were re-analysed.

Results

The additional criterion of reduced DF range in swing resulted in 45 children being classified as Type Ib and the absence of first rocker resulted in 15 children being classified as Type Ia (Table 1). These changes to the original algorithm resulted in 93% of the hemiplegic population being classified. Visual observation of the video and kinematic data of the six remaining 'unclassified' individuals revealed no deviations from normal.

Table 1. Numbers of participants classified using original and refined automated classification system

Classification	Original system	Revised system (%)
Type IV	8 (9.5%)	8 (9.5%)
Type III	6 (7.1%)	6 (7.1%)
Type II	4 (4.8%)	4 (4.8%)
Type I	31 (36.9%)	
Type Ib		45 (53.6%)
Type Ia		15 (17.9%)
Unclassified/normal gait	35 (41.7%)	6 (7.1%)

Discussion

This study has demonstrated that the classification of hemiplegic walking patterns in a total population is possible without visual confirmation. The original algorithm classified children as Type I if they failed to achieve DF>0degrees in swing, however the addition of a reduced DF range to this criteria facilitated the classification of a further 14 children. This perhaps more accurately reflects the work of Winters and Gage [2] who identified but did not define reduced DF in swing in Type I hemiplegic gait. Further modification of the system using the absence of 1st rocker (Type Ia) allowed classification of almost the entire population. The results of the application of the algorithm to a total population of children with hemiplegia suggests that many children have mild deviations from normal walking patterns that may be difficult to observe clinically.

This automated procedure can detect subtle deviations from normal and removes bias that can be difficult to eliminate from visual diagnosis. This tool may be particularly useful when evaluating large clinic or population-based samples. However it is important to note that ultimately this procedure applies discrete values to continuous data and thus some children may, over a series of different gait cycles, satisfy the criteria for inclusion in two groups. The reliability of the system in these instances has been satisfactorily evaluated and is presented elsewhere.

Acknowledgements

We would like to thank all the families that participated in the project and acknowledge the support of the Northern Ireland Research and Development Office, Project Grant RSG/1708/01.

References

[1] Kelly C, Kerr C, McDowell B, Cosgrove A. (2005) Proceedings of 14th Annual Meeting of the European Society of Movement Analysis in Adults and Children.
[2] Winters TF, Gage JR, Hicks R. (1987) The Journal of Bone and Joint Surgery 69: 437-441.

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QUANTIFICATION OF KINEMATIC MEASUREMENT VARIABILITY IN GAIT ANALYSIS

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Summary/conclusions
A method is proposed to identify the components of variability of gait analysis kinematic measures. This quantifies the repeatability of data from individual assessors and will permit identification of systematic differences between multiple assessors. Results demonstrate that individual assessors differ in error magnitude across gait parameters. Such specific information may direct clinical training and quality assurance programs.

Introduction
Repeated three-dimensional gait analysis (3DGA) of a single subject is known to exhibit inconsistency. This can be attributed to the inherent variability of walking and errors associated with the measurement process. Inconsistent marker placement has been identified as a key factor contributing to data variability across repeat sessions [1]. This variability is increased when different assessors perform the measures. Despite many reports of the reliability of 3DGA, there are limited methods available to clinical laboratories to quantify the sources of variability within kinematic data. Further, few studies (most usefully Schwartz et al.[2]) have quantifed this error in a manner that is readily accessible to gait laboratory clinical practice, identifying variation attributable to the differences between trials within a single session (inter-trial), between sessions measured by a single assessor (inter-session), and between data measured by different assessors (inter-assessor). The purpose of this study was to use the statistical analysis approach of variance components estimation to evaluate the variability of kinematic gait measures of a single unimpaired adult by six assessors from three Gait Laboratories.

Statement of clinical significance
Three-dimensional kinematic gait measurements are routinely used within clinical gait analysis to evaluate the effect of interventions on individuals. Detailed understanding of the error associated with these measures may assist clinical decision-making and provide direction to staff training and quality assurance activities within gait laboratories.

Methods
A single adult subject (22 years, height 165cm, weight 76kg) attended three Gait Laboratories for 3DGA. At each laboratory, marker placement was independently conducted by two staff members. Assessor experience ranged from novice (< 30 3DGA's) to expert (>500 3DGA's). Each assessor repeated the analysis on two occasions within the same day. For each gait session, six left and six right gait trials were captured, in a manner consistent with typical laboratory procedures. Necessary anthropometric measures obtained at the first test session were used in the subsequent session. A conventional biomechanical model (Plug-in-gait) was used to calculate lower extremity joint kinematics. Components of variability due to session and trial were estimated separately for each combination of staff member, kinematic variable and percentage of the gait cycle. A multi-level mixed-effects linear regression model was fitted with a fixed effect for side (right or left) and a random effect for session [3]. Variability due to session was estimated by the standard deviation of the session random effect and variability due to trial estimated by the standard deviation of the model's residual error term. Maximum

likelihood estimation was used in Stata 9.1 statistical software to estimate these models. The model was extended to analyse data from all points in the gait cycle simultaneously by including fixed effects for percentage point of the gait cycle.

Results
Inter-trial, inter-session and total error estimates (incorporating both trial and session variability) varied across gait parameters and assessors. As an example, Figure 1 illustrates differences in total pelvic tilt error from the 6 assessors. The magnitude of error in key gait parameters appeared related to experience, with the most experienced (ME) assessors generally recording lower errors than less experienced (LE) assessors. This was particularly evident in key parameters sensitive to marker misplacement such as pelvic tilt (ME <1deg; LE >2.5deg), hip rotation (ME <2 deg; LE >5 deg) and foot progression (ME <1.5 deg; LE >2 deg).

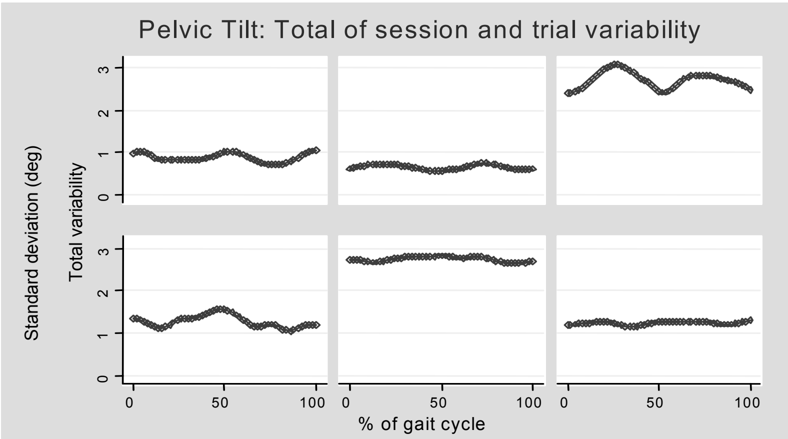

Figure 1: Errors in pelvic tilt for a single subject measured by 6 assessors. Each graph represents the total of session and trial variability from a single assessor

Discussion
Quantification of the error sources associated with 3D analysis has varied applications. Knowledge of a single assessor's typical inter-session error may be relevant in laboratories with a single staff assessor, or where a designated assessor routinely repeats 3DGA on patients returning for evaluations. Determination of inter-assessor error within a laboratory will assist in defining gait parameters which reflect systematic assessor deviations in marker placement protocols. Evaluation of inter-assessor error within and across laboratories is relevant for multi-centre collaborative research and in construction of shared gait databases. Quantitative feedback of individual and within-laboratory assessor reliability offers opportunities to more specifically direct quality assurance activities including biomechanical model review and marker placement training and practice. This method may also permit a closer evaluation of the relationships between assessor experience, training and error profiles. This report uses data from a single unimpaired subject to illustrate the potential application of this approach. The techniques are equally applicable to multiple subjects from relevant clinical populations with gait pathology.

References
[1] Gorton G et al. (2002) Gait and Posture,16 (suppl 1): p. S65-66.
[2] Schwartz MH et al. (2004), Gait and Posture, 20, 296-203.
[3] Dunn G (2004) Statistical Evaluation of Measurement Errors.

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EXTRINSIC AND INTRINSIC VARIATION IN KINEMATIC DATA FROM THE GAIT OF HEALTHY ADULT SUBJECTS

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Summary/Conclusions
3D kinematic data from 10 healthy adult subjects were collected from 3 sessions within a week. In one session, marker placement was guided by indelible pen marks made on the subjects in a data collection 2 days previously. Using this experimental design, we were able to distinguish variation due to marker placement and that due to the intrinsic variation of human walking. We found that, under near ideal circumstances, marker placement accounted for only a small amount of inter-sessional variation, with the greater proportion due to intrinsic differences in the gait patterns of the study subjects.

Introduction
3D gait analysis is used clinically to assist treatment planning and assess outcome following intervention. The ability to report changes in gait with confidence is reliant on repeatable data collection. The validity of gait analysis has been questioned (1,2) with variation in the data collected, in the interpretation of the results and in the subsequent recommendations. Previous studies have reported better repeatability in kinematic data recorded on the same day than from sessions recorded on different days, with the greater variability attributed to marker placement (3,4,5,6). In this study, we wished to quantify the variation in kinematic data due to marker placement and the intrinsic inter-sessional variability of normal walking.

Statement of clinical significance
Our results suggest when a single therapist applies retroreflective markers on subjects of moderate body-mass index, the portion of the inter-sessional standard error due to marker misplacement is less than 0.3 degrees for a set of commonly used kinematic variables. Most of the inter-sessional variation observed in pathological groups is likely to be due to intrinsic variation in human walking.

Methods
3D gait data was collected from 10 adult volunteers (mean age 31.4 years, mean BMI 22.65), with no history of musculo-skeletal abnormality, using Vicon 612 motion analysis system. A lower limb marker set was applied (modified Helen Hayes), by the same experienced therapist to each subject, on 3 separate occasions within a week. On one of the occasions, indelible ink was used to mark the skin over the anatomical locations prior to application of the marker set. These pen marks were then used to guide the marker placement for the subsequent data collection session 2 days later. One other data collection session was conducted two days before or after the marked sessions (randomised). The ink marks made on the skin were not visible during this session. In each session, data were collected during five walking trials with the subjects walking at their own self selected speed. Data were processed using the plug-in-gait model (Vicon Workstation) at which time thigh rotation was adjusted to correct the knee varus curve in swing (7). The intra- and inter- sessional standard errors (SEs) were calculated (6) for each of 9 commonly-used kinematic variables and for a combination of all these variables (Table 1). The average SE over the gait cycle was calculated for those trials belonging to a single session (intra-sessional), for those trials in marked and unmarked sessions (inter-sessional, including marker error), and for those belonging to the two marked sessions (inter-sessional, excluding marker error). Data were analysed using a two factor ANOVA (subjects, source of error). Post-hoc paired *t*-tests were conducted to establish if there were significant differences in the size of the error sources.

Results

Our results for intra-sessional and total inter-sessional error bear striking agreement to those of Schwartz *et al* (6) with total inter-sessional error being typically some 40% larger than those from within a session. The portion of inter-sessional error due to variation in marker placement is statistically significant though the difference between total inter-sessional SE and that when marker error is removed is very small with a maximum of 0.24 degrees for knee flexion.

Table 1. *Values of standard error estimated for typically-developing adult subjects. * indicates significant differences between intersessional and intrasessional error, ^ indicates significant differences between intersessional errors due to marker displacement (p<0.05).*

	Intra-sessional SE (°)	Total inter-sessional SE (°)	Inter-sessional SE excl. marker error (°)
Pelvic Tilt	0.651	0.964	0.761*
Pelvic Obliquity	0.539	0.749	0.693*
Pelvic Rotation	1.272	1.580	1.429*
Hip Flex/ext	1.092	1.541	1.368*
Hip Ab/Adduction	0.757	1.132	0.952*
Hip Rotation	1.676	2.142	2.308*
Knee Flexion	1.620	2.134	1.894*
Ankle Dorsiflexion	1.430	1.820	1.729*
Foot Progression	1.750	2.198	2.205*
Global	***1.298***	***1.813***	***1.606*^***

Discussion

Our results suggest that under ideal conditions (single therapist applying the markers, compliant subjects with moderate BMIs), the error in kinematic variables introduced by variance in marker placement is very small. The intra- and inter- sessional SE appeared to be in agreement with the work presented for 2 subjects by Schwartz (6). The errors between trials and between sessions are small, indicating good reliability in data collection where there is one person applying the markers. Inter-sessional errors, as previously reported where multiple therapists apply markers (6), may be reduced by using the same therapist to apply markers on multiple occasions. The study indicates that there is intrinsic intra- and inter-sessional variation in human walking, though for healthy adults this is small. It is likely that the increased inter-sessional variation noted in studies of pathological groups (1) is due, in the main part, to the intrinsic variability of the gait pattern of these subjects. Clinically significant inter-sessional error due to marker placement may be seen in subjects with an increased BMI, or with marked musculo-skeletal deformity, and warrants further investigation. Significant intrinsic variability may compromise the ability of gait analysis to distinguish pathological data from control data, and data taken pre- and post- intervention.

References

1. Noonan KJ(2003) Journal of Pediatric Orthopedics 279-287
2. Skaggs DL (2000) Journal of Pediatric Orthopedics 759-764
3. Kadaba MP (1989) Journal Orthpaedic Research Vol 7 849-860
4. Steinwender G (2000) Clinical Biomechanics 134-139
5. Maynard V (2003) Gait and Posture 59-67
6. Schwartz MH (2004) Gait and Posture 196-203
7. Baker R (1999) Human Movement Science 655-667

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PAIRED OUTCOME ASSESSMENT OF A MICROPROCESSOR CONTROLLED KNEE VS A MECHANICAL PROSTHESIS

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Summary/Conclusions

The study has performed a multi-factorial analysis of patient outcomes using a microprocessor controlled knee compared to a conventional mechanical knee. A paired comparison was performed on 14 transfemoral amputees. The results demonstrated an improvement in gait and balance when using the microprocessor knee which translated into increased activity during daily life.

Introduction

Microprocessor controlled knee joints appeared on the market a decade ago and have some data regarding their function. Comparative studies document improved gait symmetry[1,5] and lower energy consumption[2,3,5]. The purpose of this study was to quantify the functional characteristics of active transfemoral amputees using a microprocessor controlled knee compared to the most prescribed mechanical knee joint.

Statement of clinical significance

Recent advances in prosthetic technology are leading to increased functionality. Microprocessor controlled knee joints are more sophisticated and more expensive than mechanical mechanisms. These prosthetic components need to be objectively analyzed in order to verify the claims of manufacturers and the views of prosthetists and transfemoral amputees currently using this technology.

Methods

This study employed a crossover design whereby only the knee component was changed. Each subject was tested using a mechanical knee prosthesis (Mauch SNS or CaTech) and retested 2-3 months later after receiving a microprocessor controlled knee joint (OttoBock C-leg). Fourteen subjects were studied (mean age 42 $\pm$ 9 years). All subjects were long-term prosthesis users (20 $\pm$ 11 years). Objective gait measurements were collected with a computerized video motion analysis system utilizing ten infrared cameras (Eva RealTime 3D, Motion Analysis Corp, Santa Rosa, CA). Simultaneously, ground reaction forces were collected from four forceplates (Kistler Instrument Corp., Amherst NY, Model 8281B; AMTI, Watertown MA, Model BP2416). The 3D marker coordinates and forceplate data were used as input to a commercial software program (OrthoTrak 5.0, Motion Analysis Corp., Santa Rosa, CA) to calculate the joint kinematics and kinetics. Subjects' balance was tested on the Equitest Posturography System. Energy consumption measurements were made using a commercially available automated system (Medical Graphics Model CPX-D, St. Paul MN) modified to interface to a respiratory mass spectrometer (Perkin Elmer 1100). Testing was performed at 0.45, 0.9, and 1.3 m/sec. Total daily energy expenditure was quantified using doubly-labeled water. The Prosthetic Evaluation Questionnaire (PEQ) was used to obtained the patient's subjective perspective. Statistical comparisons were made using a paired t-test. Statistical significance was set at an alpha level of 0.05.

Results

Subjects demonstrated improved gait and balance characteristics after receiving the microprocessor controlled prosthetic knee joint. The knee moment demonstrated a significant shift from an internal flexion moment toward a more normal extension moment during midstance (Fig. 1). The subjects achieved greater gait symmetry. Their balance was significantly improved when using the microprocessor controlled knee (Fig. 2). There was no significant difference in the metabolic cost of walking when using the microprocessor controlled knee compared to the mechanical prosthetic knee. The improvements in gait and balance when using the microprocessor controlled knee resulted in a significant increase in total daily energy expenditure of 20% per day (Fig. 3). These objective findings translated into increased patient satisfaction in all aspects of their lives (Fig 4).

Fig 1. Knee Moment

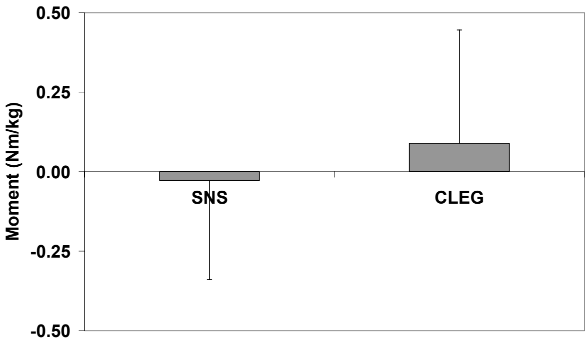


Fig 2. Sensory Organization Test

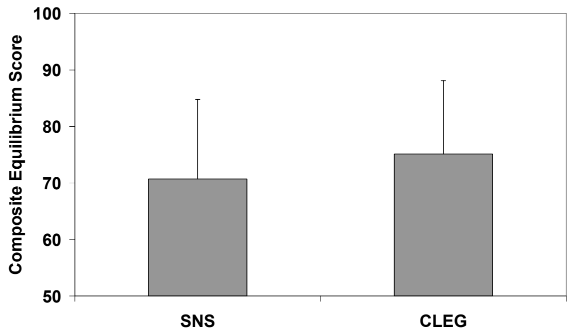


Fig 3. Total Daily Energy Expenditure

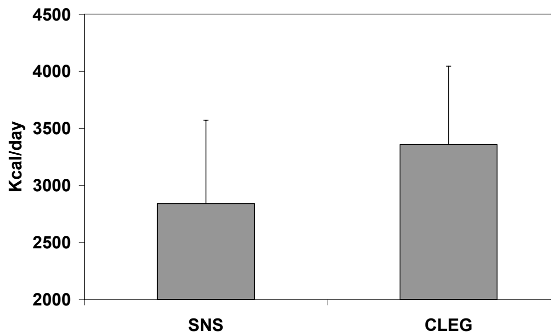
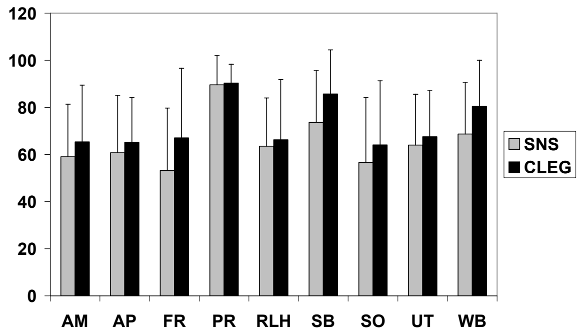


Fig 4. Prosthetic Evaluation Questionaire



Discussion

Although the first commercially available microprocessor controlled knee joint was introduced over a decade ago, less than ten studies have evaluated these devices[1-7]. Comparative studies performed on the Endolite Intelligent prosthesis, the Otto Bock C-Leg and the Ossur Rheo Knee have documented improved gait symmetry[1,5] and lower energy consumption[2,3,5]. The current study, performed on the OttoBock C-Leg, found similar results.

Individuals using a microprocessor-controlled knee demonstrate improvements in gait and balance with a concomitant decrease in energy consumption. These positive changes translate into increased activity in the users' daily life.

References

1. S. Kirker, et al., *Clin. Rehabil.,* 10:267-273, 1996.
2. M. B. Taylor, et al., *Prosthet. Orthot. Int.,* 20:116-121, 1996.
3. J. G. Buckley, et al., *Arch. Phys. Med. Rehabil.,* 78:330-330, 1997.
4. D. Datta, et al., *Prosthet Orthot Int*, 22:129-35, 1998.
5. T. Schmalz, et al., *Gait Posture*, 16:255-63, 2002.
6. J.L. Johansson, et al., *Am L Phys Med Rehibil,* 84(8):563-575, 2005.

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ENERGY EXPENDITURE DURING OVER GROUND WALKING IN PAEDIATRIC AMPUTEES

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Summary/conclusions

Results for self selected walking (SSW) in children with lower extremity amputation show that amputees of all levels tend to reduce their walking speed (a significant decrease was found in the hip disarticulation (HD) group), but the cost of ambulation and heart rate (HR) are only significantly increased in the HD group, compared to normal. Our study fails to support previous adult research that shows that as the level of amputation ascends the leg, energy cost and HR increase while SSW velocity decreases. This study demonstrates that although amputees choose to walk at a slightly slower SSW speed, the energy cost and HR of participants with Symes, below knee amputations (BKA), knee disarticulations (KD) and above knee amputations (AKA) are within normal range during SSW.

Introduction

In reviewing the amputee literature regarding energy expenditure, it is clear that the adult population is well described [1], but less information can be found on the paediatric population. Herbert et al compared children with BKA to normal during treadmill walking at self selected speed and found that patients with BKA were able to maintain similar walking speed but had significantly higher energy costs than patients with intact limbs [2]. They did not find significant differences in HR, as the adult literature reports [1]. Ashley et al. measured HR and SSW velocity in children with Symes, BKA, KD and AKA and found that all groups walked at a slightly decreased speed, with an increased HR [3].

Statement of clinical significance

It is unclear if the results reported in the adult literature can be directly related to the paediatric population at multiple levels of amputation. The purpose of this study is to compare the energy efficiency of children with lower extremity amputation between groups and to age matched normals.

Methods

Unilateral amputees over the age of five were invited to participate in this IRB approved study. Exclusion criteria for testing included: prosthetic use of less than 6 months, prosthetic complications and skin breakdown. All participants were required to refrain from eating two hours prior to testing. Oxygen consumption was collected using the K4 b^2 oxygen analysis telemetry unit (Cosmed, Rome, Italy). A five minute seated rest period was collected prior to a ten minute walk at a self selected speed around a 40 meter track. One minute of steady-state data was selected from rest and SSW, and reduced. Patients were grouped according to level of amputation: Symes, BKA, KD, AKA and HD. Each amputee was then compared to the appropriate group of age matched normals: Child 6-12yrs or Teen 13-19yrs. Variables analyzed were resting VO_2 Rate (ml/kg/min), resting HR (bpm), VO_2 Cost (ml/kg/m), HR (bpm) and velocity (m/min). A Tukey post-hoc multiple comparison procedure was run to compare amputee levels and to compare amputees with normal. The overall error rate is controlled at 0.05.

Results

Forty-three unilateral amputees and 39 normal children participated in this study. Subject demographics can be seen in Table 1. All amputees were age matched and compared to the appropriate group. The Symes group had significantly higher BMI than normal. No other differences were found in anthropometrics, or in resting VO_2 Rate or resting HR.

Table 1- Demographics of Amputees and Normal Child/Teen

	Symes n=17	BKA n=10	KD n=9	AKA n=4	HD n=3	Normal CHILD n=23	Normal TEEN n=16
Age (yr)	11.1 (±2.7)	11.2 (±4.1)	14.3 (±3.6)	15.5 (±2.6)	12.3 (±5.5)	10.0 (±1.6)	15.4 (±1.8)
BM (kg)	49.6 (±19.6)	45.6 (±23.4)	56.8 (±10.4)	66.4 (±22.1)	40.9 (±16.4)	35.6 (±7.5)	66.0 (±21.0)
BMI (kg/m^2)	*21.1 (±5.6)	19.3 (±4.9)	21.0 (±2.6)	21.1 (±4.2)	18.9 (±0.9)	16.8 (±2.1)	22.6 (±6.5)

* statistically different from age matched normals

An analysis of VO_2 Cost, walking velocity and HR was conducted comparing between amputee groups, and normal. Data is presented in Figure 1 as a percent of age matched normal. Amputee group comparisons show that AKA and HD have higher VO_2 Cost than Symes, and HD have significantly higher HR than the Symes and KD groups during SSW.

The Symes, BKA, KD and AKA groups were not statistically different than normal for VO_2 Cost, SSW velocity or HR. The HD group had significantly higher VO_2 Cost and HR while the SSW velocity was significantly slower, than normal. An overall trend can be seen in Figure 1 for increasing VO_2 Cost and decreasing SSW velocity, with amputation level.

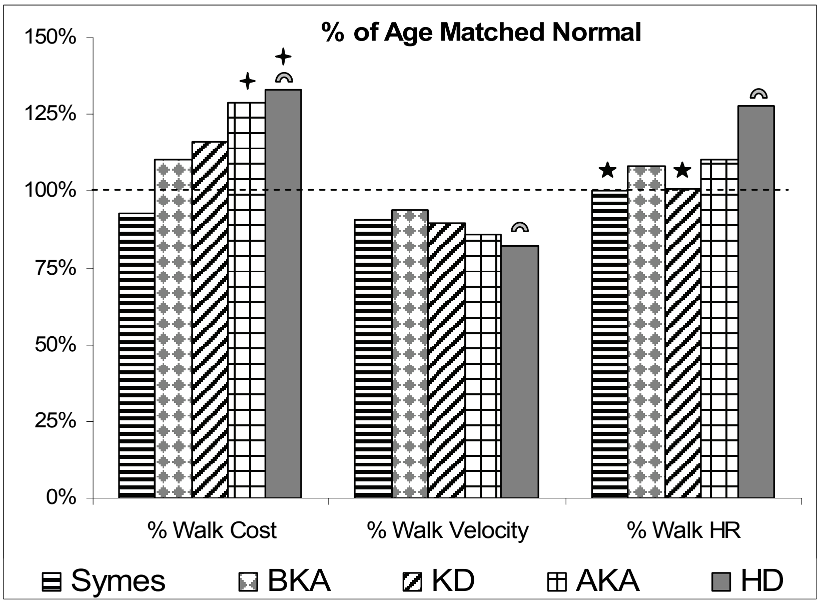

Figure 1- statistically different than: ◠ normal, ★ HD and ✚ Symes.

Discussion

The Waters series of adult amputees show that energy expenditure increases and walking speed decreases with the level of amputation for traumatic, surgical and vascular amputees. Although this study shows a trend for increased VO_2 Cost with increased level of amputation, the data shows that paediatric amputees maintain a slightly slower walking speed, but do not show a significant increase in energy cost or in HR until amputation occurs at the hip.

References

[1] Waters RL and Mulroy S, 1999, Gait and Posture, vol. 9, 207-231.
[2] Herbert LM et al., 1994, Physical Therapy, vol. 74, no. 10, 943-950.
[3] Ashley RK et al., 1992, Orthopaedic Review, vol. 21, no. 6, 745-749.

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HIGH FAILURE RATES WHEN AVOIDING UNEXPECTED OBSTACLES WHILE WALKING IN PATIENTS WITH A TRANS-TIBIAL AMPUTATION

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Conclusions

Patients with a lower leg prosthesis show significantly higher failure rates than control subjects when avoiding sudden obstacles. Under time pressure, the patients perform best when they use their non-prosthetic leg as the lead limb in a short step strategy (SSS). Some of the more experienced prosthesis users made no errors at all, which suggests that over many years it is still possible to relearn the appropriate avoidance reactions sufficiently fast.

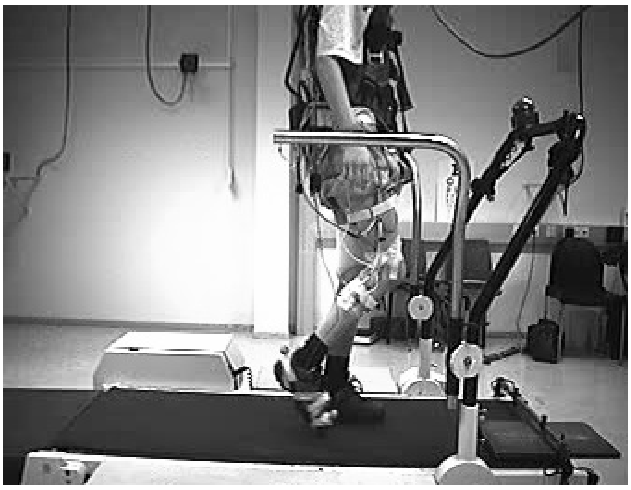

Figure 1*. Schematic diagram of experimental set-up. An electromagnet was attached to a bridge over the front of the treadmill. Two reflective markers were placed on the most posterior and most anterior end of the (right) foot and to the closest end of the obstacle.*

Introduction

Individuals with a lower leg prosthesis due to a trans-tibial amputation have no muscle control about the artificial ankle and suffer from absent propriocepsis from the ankle joint and lower leg muscles as well as from absent exterocepsis from the foot sole. Individuals with a trans-tibial amputation may experience difficulties not only with unperturbed standing and walking, but especially with avoiding obstacles [1] which increases their risk of falling. In daily life, obstacles often occur suddenly requiring fast, more or less automatic responses. In this perspective, we addressed the question if and under what circumstances patients with a trans-tibial amputation are less successful in avoiding unexpected obstacles. Because of the higher fall risk in patients with a trans-tibial amputation [1], it was hypothesized that this group of patients would indeed be less successful than a group of healthy adults when avoiding unexpected obstacles, particularly under time pressure.

Statements of clinical significance

To understand why patients with a lower leg amputation fall more easily, it is necessary to study their ability to avoid obstacles. The acquired information about obstacle avoidance strategies will be used to develop a new prosthetic training method in which reactions with both the non-prosthetic (NP) and prosthetic-leg (P) will be trained in the rehabilitation of patients with a trans-tibial amputation in order to reduce their fall risk.

Methods

Eleven patients with a trans-tibial amputation and 14 healthy controls participated in this study. Subjects walked on a treadmill at 2 km/h (Figure 1). In 2 series of 12 trials each, an obstacle was dropped in front of the P-leg or the NP-leg of the amputation group and the left leg of the control group at different phases during the step cycle. It was noted which avoidance strategy was used (a Long Step Strategy (LSS) or a Short Step Strategy (SSS)) and whether the obstacle was avoided successfully or not. These data were expressed as a percentage of the total number of trials completed by each subject.

Results

With either leg the amputation group made significantly more errors than the control subjects (24±17% and 21±17% for the P-leg and NP-leg, respectively, compared to 2±2% for the control group). Highest failure rates were found for the amputation group when time pressure was high, requiring a SSS, especially at the prosthetic side. However, a LSS under time pressure nearly always resulted in a failure for both the P and NP legs. In the amputation group, the more experienced prosthesis users were most successful in avoiding unexpected obstacles (Figure 2).

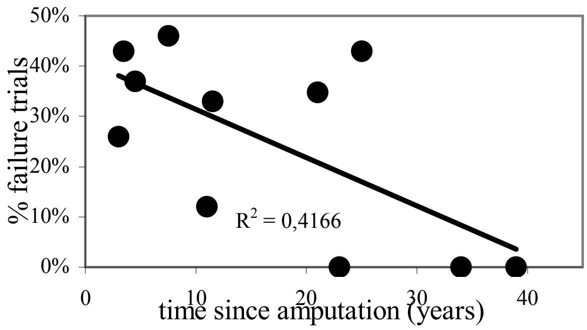

***Figure 2**. A scatter plot of the failure rate against the time since amputation for the prosthetic leg of the amputation group.*

Discussion

Patients with a lower limb amputation were very sensitive to time pressure, which was reflected in disproportionally high failure rates with either leg in the case of short reaction times (ART < 250ms). However, even with long reaction times (ART > 500ms), they contacted the obstacle more often than healthy controls. Apparently, even under low time pressure, it is difficult for patients using a leg prosthesis to adapt their stepping pattern to avoid the obstacle successfully, irrespective of the leg used to cross the obstacle. Remarkably, no errors were made by 3 of the 5 persons who had used a lower leg prosthesis for more than 20 years after a traumatic trans-tibial amputation. In general, there was a substantial association between failure rate of the P-leg and time since amputation, yielding an explained variance of more than 40%. These findings suggest that it is still possible to relearn the appropriate avoidance reactions sufficiently fast although this may take many years and may only be true for otherwise healthy subjects.

References

[1] Miller WC, Speechley M, and Deathe B. 2001. The prevalence and risk factors of falling and fear of falling among lower extremity amputees. Arch Phys Med Rehabil, 82: 1031-1037.

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SUBJECT SPECIFIC HIP GEOMETRY AFTER THP INFLUENCES HIP JOINT REACTION FORCES DURING GAIT

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Summary/conclusions
Subject specific hip geometry (especially NSA) after total hip joint replacement alters the joint reaction forces at the hip and affects implant loading.

Introduction
Each year, 17 000 hip prostheses are implanted in Belgium. 5 – 10% need revision because of implant loosening. Several factors have been identified to predispose implant loosening. Musculoskeletal loading is often reported as an important factor affecting the biological processes involved in bone remodeling and primary fixation of implants. In patients with THP, the subject specific anatomy of the hip implant determines the moment generating capacity of the surrounding muscles. Our previous work [3] reported a decrease in moment generating capacity of the hip abductors when a subject specific model was used that incorporates RX based values for femoral neck length (NL), femoral neck-shaft angle (NSA) and width of the pelvis (WP). As a result, we reported marked changes in calculated muscle (co-) activations using a static optimization approach. The present work evaluates to what extent hip contact forces during gait are influenced by subject-specific geometry of the hip and the resulting changes in muscle activation balance during gait.

Statement of clinical significance
A better insight in the biomechanical factors influencing hip loading during gait after THP will contribute to an enhanced understanding of the factors affecting initial implant fixation and eventually prevent implant loosening.

Methods
NL, NSA and WP were measured in 20 subjects, based on digitized post-operative RX-images (Imagica, GreyStone Inc). A deformable musculoskeletal model of the lower limb was adjusted to incorporate NL, NSA and WP for all patients (SIMM, Musculographics). NL varied from 41 mm to 86 mm, NSA between113° and 144°, PW between 315 mm and 402 mm. Kinematic and kinetic data obtained from a normal gait trial were imposed to each individualised model to calculate (1) joint moments (2) individual muscle force generating capacity and (3) muscle moment arms over the gait cycle. Muscle activation patterns balancing the external joint moments were computed using a static optimisation algorithm, minimizing the sum of the muscle forces (Matlab, MathWorks Inc.). The 3D hip reaction forces were computed taking into account the muscle forces resulting from the muscle activation patterns, as well as external forces (the ground reaction forces – inertial forces and gravity). This analysis was repeated for a model with halve hip abductor force generating capacity, mimicking hip abductor weakness after surgery. The resulting changes in muscle activations were imposed[1]. Peak reaction forces during single limb stance as well as the associated inclination of the reaction force in the sagittal and the frontal plane are reported.

Results
The peak reation forces in the undeformed model are within the range reported in literature [2]. The changes in muscle activations due to the modified geometry introduced changes in peak reaction forces (Figure 1). Consistent changes were most pronounced for the mediolateral component showing a decrease in the peak mediolateral component with increasing NSA.

A distinct trend to increased peak vertical joint reaction forces with increased NSA was found. As a result, the inclination angle of the resultant reaction force in the frontal plane decreased substantially with decreasing NSA, introducing a more vertical alignment of the reaction force at peak loading (Figure 2). Although, the sagittal plane angle showed variations in the individual model, no clear relation with the changes in NSA could be established. When halving the abductor muscle force generating capacity, all components of the reaction forces decreased. However, the previously described relations of the joint reaction forces with NSA were unchanged.

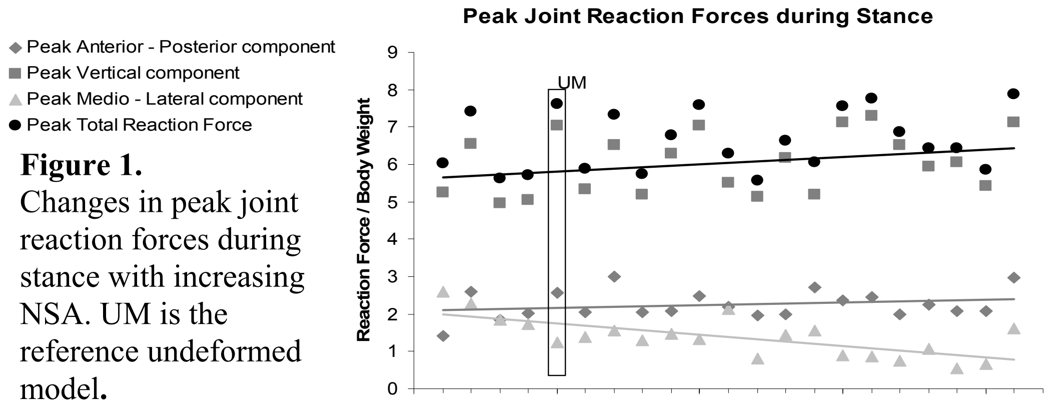

Figure 1. Changes in peak joint reaction forces during stance with increasing NSA. UM is the reference undeformed model.

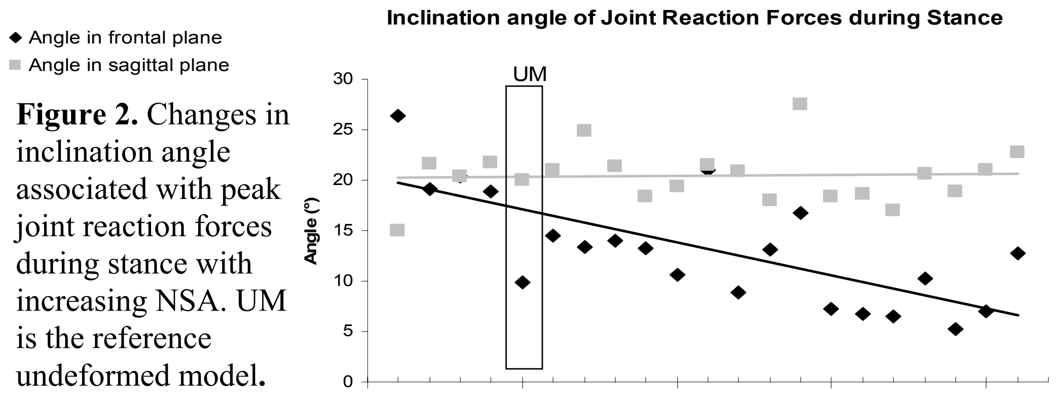

Figure 2. Changes in inclination angle associated with peak joint reaction forces during stance with increasing NSA. UM is the reference undeformed model.

Discussion

The interaction between hip geometry, muscle moment generating capacity and hip loading can be analyzed using inverse dynamics based on personalized musculoskeletal modeling.
Our results show an important interaction between NSA and medio-lateral reaction forces, causing a more vertical inclination angle in the frontal plane during gait. This therefore will result in a more vertical loading of the implant when peak reaction forces are applied during gait. Although we did not analyze the interaction between NSA, NL and PW, our findings do suggest that a minimal NSA angle needs to be preserved in order to limit the medio-lateral force component and limit the resulting bending stress in the femoral shaft. The calculation of individual hip loading by use of musculoskeletal modeling and inverse analysis may contribute to understanding the effect of hip joint loading on bone remodeling and implant load shearing.

References

[1] Lenaerts G, Jonkers I, Spaepen A, 2005, Gait & Posture, 22 (1), S19:
[2] Stansfield BW, Nicol AC, 2002, Clin Biomech, 17:130-139

O-38

BIOMECHANICAL AND METABOLIC PERFORMANCE OF A NEW ORTHOTIC KNEE JOINT PRINCIPLE FOR STANCE PHASE SAFETY

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Summary

Recently developed orthotic knee joints that are integrated into a knee-ankle-foot-orthosis (KAFO) stabilize the paralyzed knee in stance phase while permitting natural flexion and extension at the knee during swing phase. The functional advantages for these patients can be documented by metabolic and biomechanical parameters.

Introduction

For many patients with lower limb weakness or paralysis, functional mobility is achieved by means of a KAFO. In traditional orthoses, the knee joint is locked throughout the complete gait cycle (knee completely locked – KTL) to guarantee safety during stance phase. Over the last few years, 5 new orthotic systems ([1]) have been developed worldwide which stabilize the knee joint during stance phase without interfering with knee flexion in swing (knee locked during stance – KLDS). In the present study, KAFOs utilizing KTL and KLDS principles are compared based on metabolic and biomechanical parameters.

Statement of clinical significance

This study investigates and documents the functional advantages that patients with lower limb weakness or paralysis can anticipate from orthoses utilizing the new KLDS principles.

Methods

6 patients (5 male patients and 1 female patient, aged 48±14y, height 177±9cm, body mass 83 ±14kg) with paresis of the m.quadr. fem. who had been fitted with a KAFO (Free Walk, Otto Bock, Germany) were examined. The knee joint in this orthosis can be used in either KTL and KLDS settings. Metabolic energy consumption was measured while walking on a treadmill (velocities between 0.6 and 0.95m/s) using the CPX system (MedGraphics, USA). Kinematic and kinetic parameters determined while walking on level ground at each subject's self-selected comfortable speed (VICON 460, Oxford Metrics, GB; Kistler Force Plates, Kistler AG, Switzerland). A similar gait analysis on level ground was conducted with a group of healthy people (n=30, age 28±5y, height 175±6cm, body mass 71±9kg) for comparison.

Results

Using the KLDS principle on a treadmill, a significant reduction of oxygen consumption averaging 12% was demonstrated. Average heart rate also dropped significantly by 7% (table 1). Biomechanical gait parameters demonstrated that the abnormal pelvic obliquity, necessary to clear the foot when the KAFO was in the KTL mode, is eliminated in the KLDS mode. (Figure1, left). Furthermore, the distinct overload on the joints of the contralateral leg, which was present in KTL mode, is significantly reduced in the KLDS situation. (reduction of strong external extension moments in knee and hip joint, figure 1, right).

Table 1: Mean O_2 rate, mean O_2 cost and mean heart rate for the measurements with KTL and KLDS (**, *: significance with p<= 0.01, 0.05 according to the WILCOXON Test)

Parameter:	KTL	KLDS	Difference KTL-KLDS [%]	Significance
O_2 rate [ml/min*kg]	13.9+/-1.9	12.4+/-2.0	12	**
O_2 cost [ml/kg*m]	0.362+/-0.09	0.324+/-0.089	11	**
Heart rate [min^{-1}]	110+/-12	103+/-12	7	*

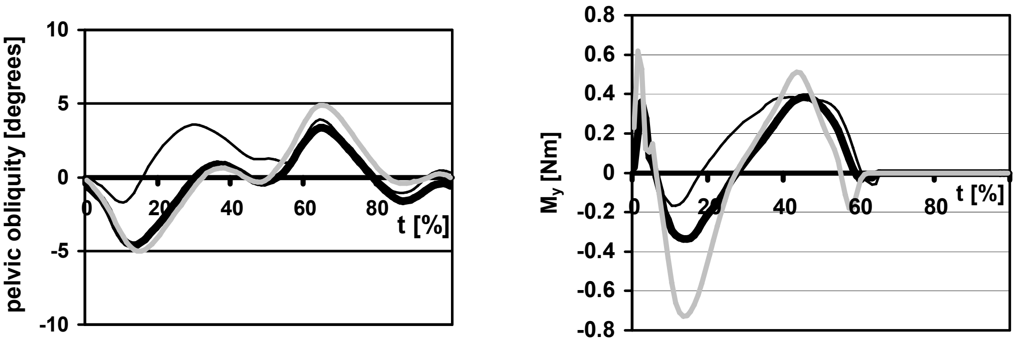

Figure 1: Mean biomechanical parameters for KTL (thin black), KLDS (thick black) and the controls (grey; left: pelvic obliquity for the cycle of the contralateral leg, right: external stance phase knee moment of the contralateral leg measured in the sagittal plane)

Discussion

The results of this investigation of metabolic parameters demonstrate that the KLDS principle provides significant advantages to the patient with lower limb weakness or paralysis. The biomechanical reason appears to be the elimination of the compensatory vertical movement of the Center of Gravity that is necessary when the knee joint is fixed in extension during swing phase. ([2], [3]). In the KLDS mode, extension moments of the contralateral leg were reduced to a nearly normal level.

References

[1] Schmalz, T. Blumentritt, S., Drewitz, H.: Gangphasenabhängig entriegelnde versus gesperrte Beinorthesen – Biomechanische und metabolische Untersuchungen. Med.Orthop.Tech. 125 (2005) 3, 67-74.

[2] Mattsson, E., Broström, L.: The increase in energy cost of walking with an immobilized knee or an unstable ankle. Scand.J.Rehab.Med. 22 (1990), 51-53.

[3] Kerrigan, C., Viramontes, B., Corcoran, P., Laraia, P.: Measured versus predicted vertical displacement of the sacrum as a tool to measure biomechanical gait performance. Am.J.Phys.Med.Rehab. 74 (1995) 1,3-7.

O-39

BIOMECHANICS OF GAIT IN CHARCOT-MARIE-TOOTH DISEASE

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Summary/conclusions
Motion data indicate that there are distinctively different types of gait patterns in children and youth with CMT. The primary deformity seen in most cases was late peak ankle dorsiflexion in stance (ST); the secondary deformity was excessive peak ankle dorsiflexion in ST. Excessive equinus in swing or drop foot was much less common. The variation in gait patterns would suggest the need for different treatment strategies spcific to the individual and depending on gait presentation.

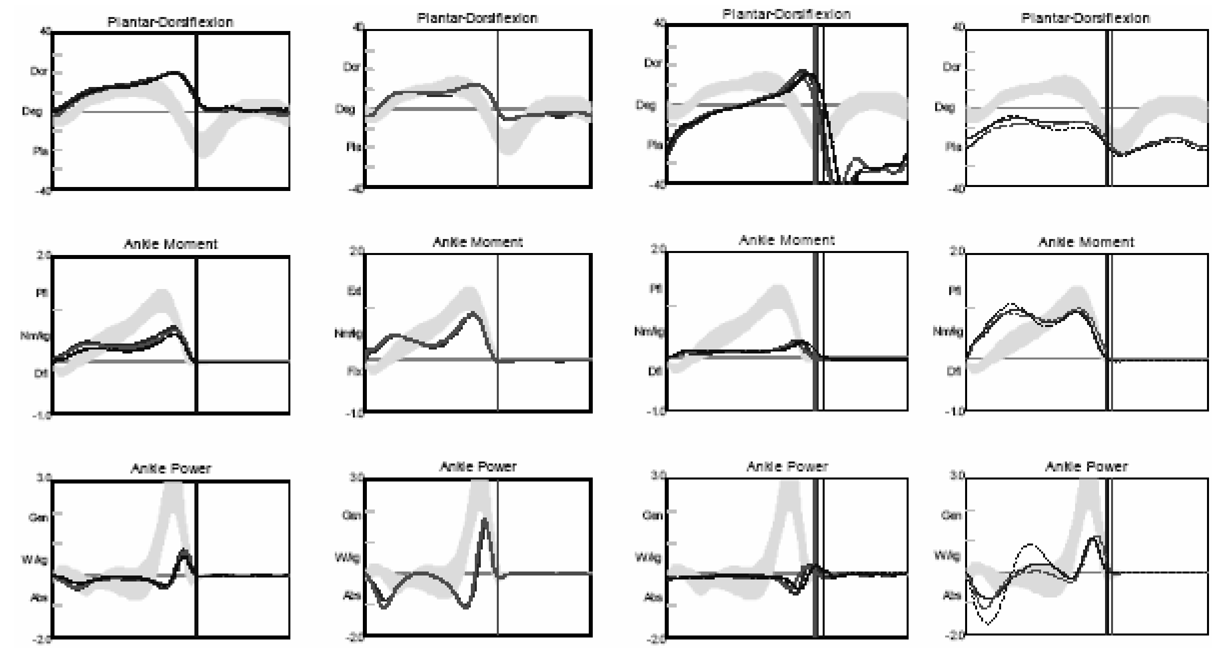
Introduction
Charcot-Marie Tooth (CMT) is characterized by distal muscle weakness and imbalance with associated gait implications which progress at varying rates. The textbook gait description includes: foot drop in swing phase, steppage (hyper-flexion of knee and hip in swing), circumduction and pelvic hiking. However, clinical observation would suggest that gait patterns in persons with CMT do not match a single set of parameters. It is possible that there are a variety of patterns of gait in CMT that could be defined more accurately with the use of computerized motion analysis. Therefore, the purpose of this study was to document objectively the characteristic gait patterns in children with CMT.

Statement of clinical significance
Identification of gait patterns in CMT would improve our understanding of the pathomechanics of gait in persons with CMT. This would also provide a stronger basis for determining the prognosis for future ambulatory function, treatment decisions and ultimately improve treatment outcomes.

Methods
Nineteen patients (9 male/10 female) with a mean age of 13±3 (range 6 to 19) years and a diagnosis of CMT where included in this study. All patients were independent ambulators with no previous lower extremity surgeries. All had ankle/foot deformity for which surgical intervention was being considered. Each patient underwent a three-dimensional motion analysis using standard techniques[1]. A comprehensive clinical examination including passive joint range of motion and muscle strength measures was also completed. All gait data were compared to normal reference data collected in the same laboratory[2]. A Students t-test was used for comparisons with normal reference data ($p<0.05$).

Results
Clinical Findings: Large subject-to-subject variations in strength and passive range of motion (ROM) were noted. Passive ROM measures were normal for ankle plantar flexion, forefoot inversion and eversion. Mean ankle dorsiflexion was consistently less than normal at 1±6º (knee 0º) and 5±8º (knee 90º). Muscle strength measures varied from normal to absent for the ankle dorsiflexors with the mean of 3±2. The foot deformity during standing included the following (out of 38 sides): 33 cavus deformity, 29 claw toe deformity, 4 hindfoot varus, 4 hindfoot valgus, 15 plantar flexor contracture (knee at 0º) and 5 plantar flexor contracture (knee at 90º).
Gait: Variations in ankle kinematic and kinetic patterns were noted across subjects (Figure 1). The most common gait deviation was delayed peak dorsiflexion in ST (34 of 38 sides). The second most common was increased peak dorsiflexion in ST (20 of 38 sides). Excessive

equinus in terminal swing (SW) was noted for 13 of 38 sides and consistent with dorsiflexor weakness. Variations in the degree of dorsiflexion in ST were due to variations in ankle plantar flexor strength and/or presence of plantar flexor contracture. The mean peak ankle plantar flexor moment 0.91±0.36 N.m/kg (normal = 1.2±0.2 N.m/kg) and power 2.23±0.54 W/kg (normal = 3.3±1.0 W/kg) were significantly less than normal consistent with ankle plantar flexor weakness noted in 32 of 38 ankles.



a) b) c) d)

Figure 1. Comparison of sagittal plane ankle kinematic and kinetic patterns for representative individuals with CMT. The most common pattern a) normal ankle dorsiflexion in SW and delayed and increased peak dorsiflexion in terminal ST - 13 sides, b) normal ankle dorsiflexion in SW and delayed but normal peak dorsiflexion in ST - 10 sides, c) excessive ankle plantar flexion in ST and increased and delayed peak dorsiflexion in ST - 6 sides and d) excessive plantar flexion in ST and SW - 4 sides.

Discussion

Gait analysis data indicate that persons with CMT demonstrate a variety of functional presentations. Gait patterns were not age related which highlights the variable penetrance of this genetic disorder. Although the textbook description reports a foot drop due to anterior tibialis weakness, a significant number of our patients presented with plantar flexor weakness and associated prolonged dorsiflexion or delayed heel rise with an absence of foot drop. Gait analysis data that demonstrates ankle plantar flexor weakness may be the first and most common functional sign that CMT may be present. The above findings would not support the naming of this disease as a peroneal nerve palsy and indicate that treatment strategies need to be specific to the actual functional deficits, which are unique to each patient.

References

[1] Davis RB, (1996), Human Motion Analysis: Current Applications and Future Directions, 17-42.
[2] Õunpuu S, (1991), J Pediatric Orthopaedics, 11, 341-349.

O-40

SAGITTAL PLANE LOWER EXTREMITY KINEMATICS IN CHILDREN WITH HEREDITARY SPASTIC PARAPLEGIA

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Summary/conclusions
Children with Hereditary Spastic Paraplegia (HSP) demonstrate one of four sagittal plane gait patterns distinct from children with spastic diplegic cerebral palsy as revealed by computerized gait analysis of the lower extremities. Further analysis with additional subjects and upper extremity kinematics will substantiate these patterns.

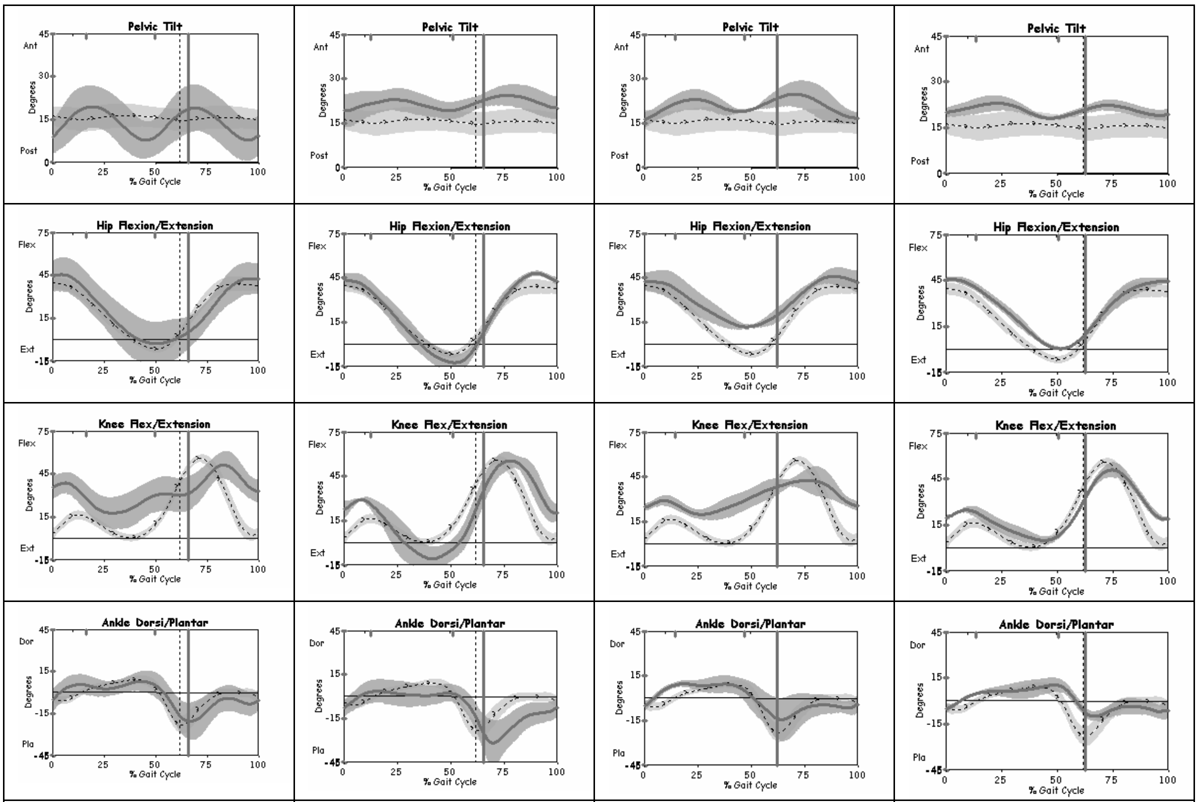
Introduction
Hereditary Spastic Paraplegia refers to a group of more than 30 inherited neurologic disorders in which the predominant characteristic is lower extremity spasticity resulting in an atypical gait pattern [1]. Since the first clinical descriptions of HSP by Strümpell in 1880, few studies have been published describing the gait patterns of HSP [2] and to date minimal kinematic information exists for the pediatric population.

Statement of clinical significance
Defining sagittal plane gait patterns specific to children with HSP will allow the clinician to differentiate between this population and children with spastic diplegic cerebral palsy, better directing diagnosis and treatment. The use of computerized gait analysis for children with HSP reveals patterns that cannot be distinguished by visual gait inspection.

Methods
A retrospective chart review was performed for 18 patients (10 female, 8 male; mean age 12.7 yrs, range 3.9 to 18.7 yrs) with HSP who underwent routine clinical gait analysis. The diagnosis of HSP was confirmed by family history (14 patients) or clinically confirmed by a neurologist (4 patients). Patients included were independent ambulators without orthoses or prior surgical intervention. Computerized gait analysis was performed using a VICON motion system. Patient gait trials were processed using Vicon Clinical Manager. One representative trial for each of the right and left sides was chosen for analysis. Because it was noted that the patients had symmetric patterns, a representative side was chosen for illustration. Utilizing the gait classification system described by Rodda [3], six clinicians experienced in gait analysis attempted to classify the HSP patients into the diplegic gait patterns. As 11/18 HSP patients did not fit into any of these categories, the purpose of this study was to determine whether patterns specific to HSP could be differentiated.

Results
Sagittal plane kinematic profiles of HSP patients grouped into four patterns where the primary differences were at the knee and pelvis (fig.1). Ankle position did not present as a distinguishing characteristic. Group A consists of 6 patients (2 female, 4 male) with a mean age of 11.4 ± 2.7 yrs and is characterized by an exaggerated double-bump pattern at the pelvis and a triple-bump pattern at the knee (an additional flexion peak in late stance). Group B includes 4 patients (all female, mean age 5.4 ± 2.2 yrs) with primary characteristics of increased peak knee flexion in loading and swing as well as knee hyperextension in midstance. Group C consists of 4 patients (1 female, 3 male) with a mean age of 17.7 ± 1.0 yrs. This group has a moderate double bump pelvic pattern in anterior tilt, hip flexion in terminal stance, a crouched/stiff knee pattern and peak ankle dorsiflexion near norms. Group D includes 4 patients (3 female, 1 male) with a mean age of 17.0 ± 2.4 yrs. This group is the mildest pattern with a slight double-bump pelvic pattern, hip extension to neutral in terminal stance and knee

flexion greater than norms in terminal swing but close to norms in terminal stance. Ankle patterns in all groups lack 1st rocker, have early dorsiflexion following initial contact and have slight dropfoot in terminal swing.



Group A	Group B	Group C	Group D

Figure 1. *Sagittal plane kinematic pattern ensemble averages ± 1s.d. for groups A-D with normal data ± 1 s.d. plotted for comparison.*

Discussion

The exaggerated pelvic tilt pattern of group A may reflect the increased trunk movement observed frequently in patients with HSP. While the triple-bump knee pattern is not immediately apparent with observational assessment, trunk movement frequently is obvious. The incorporation of upper extremity data would provide insight into the role of the trunk and arms during gait in children with HSP. Group B has a young mean age (5.4 ± 2.2yrs) possibly reflecting a lower level of spasticity or contracture development. Group C has similarities to the crouch gait pattern of spastic diplegia in the flexed hip and knee position, but lacks both the posterior pelvic tilt and ankle hyperdorsiflexion typically seen in true crouch gait. It is noted that group D includes three of the four patients without family history of HSP. Further analysis with additional patients will further substantiate these patterns.

References

[1] Fink JK, (2003), Arch of Neurol, 60:1045-1049.
[2] Klebe S et al, (2004), J of Neurol, 251:571-578.
[3] Rodda JM et al, (2004), JBJS (Br), 86-B (2):251-258.

Acknowledgements

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O-41

AN EXPLORATION OF THE FUNCTION OF THE TRICEPS SURAE DURING NORMAL GAIT USING FUNCTIONAL ELECTRICAL STIMULATION

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Summary/conclusions
This study uses random bursts of functional electrical stimulation (FES) to perturb normal gait. The results are used as an indicator of the normal function of the calf muscles and also to provide supporting evidence for predictions based on induced acceleration analysis (IAA). Gastrocnemius and soleus are shown to have opposing actions at the knee and ankle during second rocker. These counterintuitive results correlate with published predictions from IAA.

Introduction
The triceps surae muscle comprises soleus and the two heads of gastrocnemius. All cross the ankle and subtalar joints, having plantarflexing moment arms. Only gastrocnemius crosses the knee, where it has a flexing moment arm. Manual muscle testing produces a predictable response and, intuitively, the common tendon suggests a similar action at the ankle.
Computer simulation techniques have been used to investigate the role of particular muscles during gait. Recent work using IAA has yielded some surprising results for the calf muscles, demonstrating that gastrocnemius and soleus sometimes have opposing functions [1] [2]. IAA models, results, presentation and interpretation vary and the approach is not without its critics [3]. There is a clear need for additional practical validation.
The aim of this study was to investigate the dynamic action of the triceps surae muscles, during normal gait, using discrete bursts of FES. IAA can also be considered to be a perturbation away from the existing dynamics, hence the FES results can be compared directly with IAA predictions.

Statement of clinical significance
The triceps surae muscles are frequently implicated in the development of pathological gait patterns in conditions such as cerebral palsy. Many different treatments are recommended, including surgery, Botulinum toxin injection and stretching programmes. In order to treat the muscles appropriately it is important to understand their function during normal walking.
Computer simulation is becoming a clinically important tool in the decision making process following gait analysis. If a direct link can be demonstrated between the response of subjects in the gait laboratory and the results of the models then researchers and clinicians will have much greater confidence in using this kind of information.

Methods
5 adult male subjects volunteered to take part in the study. No subject had any pathology affecting their walking. Each had one lower limb tested, selected at random. One pair of FES electrodes was placed over the lateral head of the gastrocnemius muscle and a second over soleus. The timing of the stimulation was controlled using foot switches. The pulse width of the stimulation was altered to elicit as strong a contraction as was tolerable.
A full body marker set was used and gait data collected using a Vicon 612 system in conjunction with a single Kistler force plate. Six different stimulation conditions were tested, the two muscles with three different timings - first, second and third rockers. The order of the testing was randomized. In each case a series of walking trials was collected, with each trial randomly assigned for stimulation or no stimulation. For the stimulation trials only the stance

phase on the force platform was stimulated. Data collection continued until at least 6 good trials had been collected with and without the stimulus.
The means of the stimulated and unstimulated trials were compared. The difference showed the effect of the FES perturbation and was interpreted as the dynamic action of the muscle. For example where the stimulation caused greater flexion the action was defined as ‘flexing’.

Results
The results for the knee and ankle kinematics during second rocker are shown [Figure 1]. This was the period in which difference between the muscle actions was most prominent.

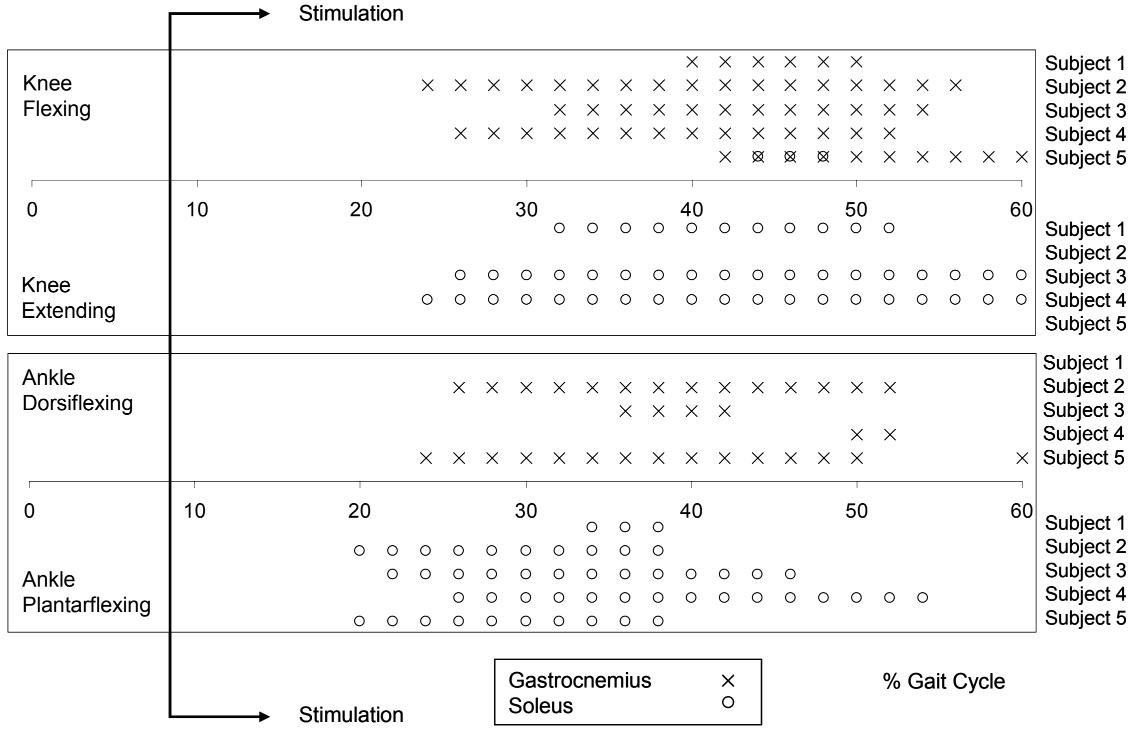

Figure 1. *Action of gastrocnemius and soleus during second rocker. A point is marked on the graph when the mean of 5 stimulated traces differed from the mean of 5 interspersed unstimulated traces by at least 2 standard deviations.*

Discussion
The responses of individual subjects to the stimulation varied, however a clear pattern emerges. During second rocker gastrocnemius and soleus have antagonistic actions. The action of gastrocnemius as an *ankle dorsiflexor* and knee flexor is certainly surprising. These results confirm predictions from previous computer model simulations, in that they highlight clear differences in action between two components of the triceps surae [1] [2]. The action at the knee is the same as that reported by Neptune et al. (2001). The next step is to perform subject-specific IAA simulations based on the data collected and compare the effects of FES perturbations with the IAA predictions in each case. It will be interesting to see whether the IAA results confirm the patterns shown above and reflect the same inter-subject variability.

References
[1] Hof AL & Otten E, (2005), *Gait Posture*, 22:182-188.
[2] Neptune RR, Kautz SA & Zajac FE, (2001), *J Biomech*, 34:1387-1398
[3] Chen G, (2006), *Gait Posture*, 23:37-44

DYNAMIC SIMULATIONS OF SLOW, FREE, AND FAST WALKING: HOW DO MUSCLES MODULATE FORWARD PROGRESSION?

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Summary/conclusions

Forward progression during walking is modulated by the same muscles, regardless of speed. Propulsion from hip flexors and plantarflexors increases with speed, as does the braking actions of knee extensors.

Introduction

Unimpaired children can increase or decrease walking speed with ease, a task that is often difficult for children with musculoskeletal disorders. However, the mechanisms by which unimpaired individuals modulate walking speed are unclear. Prior EMG studies have shown that muscle activity increases with walking speed (e.g., [1]), but how this altered activity influences walking dynamics is unknown. Muscle-actuated simulations have provided insight into how muscles generate propulsion during walking at a typical free speed (e.g., [2-3]). Using subject-specific dynamic simulations of walking, we tested the hypothesis that faster walking speed is achieved by increasing the propulsive actions of the same muscles that modulate progression at a normal walking speed.

Statement of clinical significance

A common therapeutic goal for patients with impairments is to improve walking speed. Understanding how muscles generate forward progression at slow and fast speeds in unimpaired children may aid the development of more effective treatments to improve walking speed in patients with gait abnormalities.

Methods

Three-dimensional kinematics and ground reaction forces for an unimpaired child (11 y.o) were collected at self-selected slow, free, and fast overground walking speeds. We scaled a musculoskeletal model (21 degrees of freedom, 92 muscle-tendon actuators) to the subject's anthropometry and used computed muscle control [4] to generate forward dynamic walking simulations at each speed. Computed kinematics and muscle excitations were consistent with experimental kinematics and EMG. Each simulation consisted of a half-gait cycle, beginning at the onset of double support (loading response of left limb, preswing of right limb). Muscle-induced fore-aft accelerations of the body mass center were computed using a perturbation analysis [3]. Accelerations from stance-limb muscles were averaged over four phases: loading response (double support, negative fore-aft GRF), early single-limb support (negative fore-aft GRF), late single-limb support (positive fore-aft GRF), and preswing (double support, positive fore-aft GRF). In each phase, muscles that generated ≥ 80% of the total braking or propulsive induced acceleration were identified.

Results

At all speeds, the net influence of stance-side muscles was to slow forward progression during loading response and early single-limb support, and to propel the body forward during late single-limb support and preswing (Fig. 1). Braking and propulsion by stance-side muscles increased with walking speed. Individual muscle contributions to progression reflected this trend (Fig. 2).

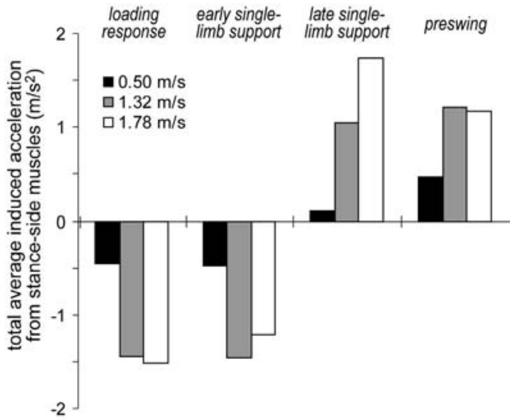

Figure 1. *Average fore-aft induced accelerations from stance-side muscles.*

Discussion

We generated subject-specific, muscle-actuated, 3D simulations of a child walking at three speeds. Increased walking speed arose from greater propulsion by hip flexors and ankle plantarflexors during late single-limb support and preswing. A counterintuitive finding was that knee extensors generated larger braking effects at higher speeds. The braking effects of dorsiflexors at the slowest and fastest walking speed were similar, suggesting unusually high dorsiflexor forces during slow walking. This finding is consistent with a prolonged internal dorsiflexion moment computed from the subject's experimental data during slow walking.

These results suggest that therapies aimed at increasing hip flexor and plantarflexor strength may be beneficial for patients with limited walking speed, since these two muscle groups are primarily responsible for increasing gait speed.

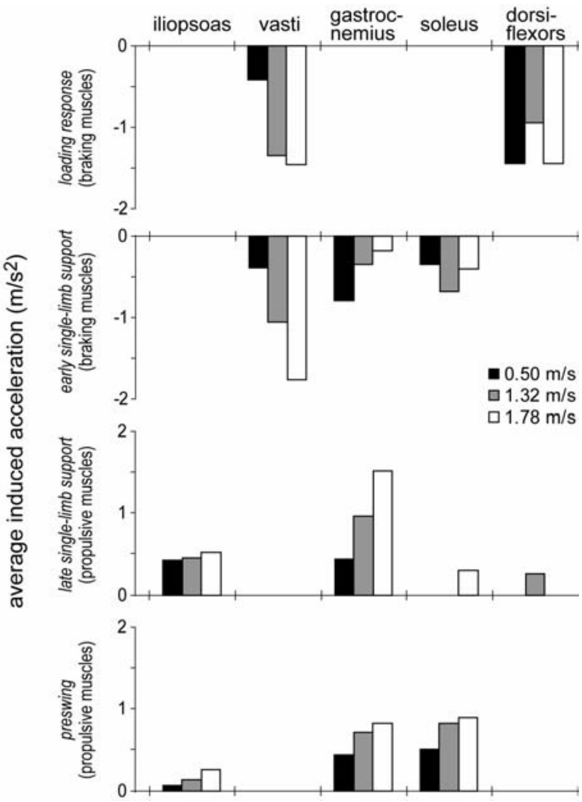

Figure 2. *Major muscle contributors to braking and propulsion. Top panel: Vasti and dorsiflexors were the primary sources of braking during loading response. Second panel: Vasti, gastrocnemius, and soleus provided braking during early single-limb support, with some assistance from rectus femoris (not shown). Third panel: Gastrocnemius and iliopsoas propelled the body forward during late single-limb support. Smaller contributions from gluteus minimus, medial hamstrings, peroneal muscles (not shown), soleus, and dorsiflexors also assisted in propulsion, but were inconsistent across walking speeds. Bottom panel: Gastrocnemius, soleus, and iliopsoas provided propulsion during preswing, with smaller contributions from medial hamstrings and peroneal muscles that were inconsistent across walking speeds.*

References

[1] den Otter, et al, (2004), Gait Posture, 19, 270-278
[2] Neptune et al, (2004), Gait Posture, 19, 194-205
[3] Liu et al, (in press), J Biomech
[4] Thelen and Anderson, (2006), J Biomech, 39, 6, 1107-1115

O-43

AUTOMATIC IDENTIFICATION OF MUSCLE INSERTION SITES IN MR IMAGES USING ATLAS-BASED, NON-RIGID REGISTRATION

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Summary/conclusions
Atlas-based non-rigid image registration allows automatic identification of muscle attachments that can be incorporated in subject-specific musculoskeletal models applicable to the biomechanical analysis of gait.

Introduction
Gait analysis techniques have evolved from research instruments to an indispensable tool in the management of patients suffering from a wide variety of medical conditions. Recent studies show its added value over clinical examination data in selecting a patient-tailored surgical intervention strategy [1]. More recent, newly 'derived' parameters, related to musculoskeletal geometry (e.g. muscle length, muscle moment arms) as well as muscle–tendon dynamics (e.g. force generating capacity of muscles), have been introduced in the clinical decision making process. Consequently, generic biomechanical models used to date need to be accommodated for inter-individual variability in musculoskeletal geometry [2], especially when bony deformations and changes in muscle geometry are present.
Highly accurate, subject-specific musculoskeletal models can be build using information of magnetic resonance (MR) images [3]. Since manual delineation of soft-tissue structures is required, parameterization of the muscle geometry, with identification of muscle paths from origin to insertion, is a time demanding step. Therefore, this approach is generally considered to be too labour-intensive and high-priced.
This work presents the performance of a method to automatically identify muscle geometry (attachments and muscle path) by atlas-based non-rigid image registration. This approach decreases considerably the time required to build subject specific musculoskeletal models based on medical images (MRI) while demonstrating an accuracy comparable to manual delineation.

Statement of clinical significance
Automatic identification of muscle geometry based on medical imaging enhances the feasibility to build subject specific musculoskeletal models for detailed biomechanical analysis of gait deviations in patients.

Methods
The current work assumes that muscles can be modelled as a straight line running from origin to insertion, describing the muscle's lines of action. This approach is based on the muscle model used in SIMM (Software for Interactive Musculoskeletal Modelling, Musculographics Inc.). Therefore, the identification of the attachment and insertion sites of the muscle is a crucial step in the process for which mostly a centroid approach is used.
Via non-rigid registration, muscle attachment and insertion sites are identified in T1 MR images of a subject using the information of an atlas (i.e. a MR images from a non-pathologic adult man (25y) with labelled muscle attachments). In a first step the geometrical relation between both image volumes is determined by matching both image volumes using intensity-based non-rigid image registration. The non-rigid deformation of the atlas image is modelled by a B-spline deformation mesh [4]. The cost function uses mutual information [5] as a

similarity measure while a volume penalty term discourages improbable or impossible deformations. In a final step, the positions of the patient's muscle origins in the respective image volume are retrieved after applying the resulting 3D deformation field to the original atlas coordinates.

Results

The muscle attachments of 8 major hip muscles were manually identified in T1 weighted MR images from a 23 year old, non-pathologic male subject. These were compared to the locations identified using the atlas-based non-rigid registration. Absolute distances between the automatically detected muscle origins and their manually defined positions were calculated [table 1]. In addition, the muscle paths were further delineated to quantify the difference between both approaches for the calculation of moment arm lengths at the hip in adduction, flexion, and rotation [table 1]. The centre of the hip joint was automatically detected by fitting a sphere to the segmented femoral head using the ICP algorithm [7].

Table 1: Comparison between manual and automatic detection of muscle origins.

	Absolute distance between automatically and manually defined muscle origin (mm)	Difference in moment arm length with respect to the hip joint (mm)		
		Adduction	Rotation	Flexion
Adductor longus	31.1	24.7	0.2	0.2
Adductor brevis	15.9	11.1	0.7	3.4
Rectus femoris	5.9	4.5	1.21	1.0
Gracilis	4.4	3.9	0.9	1.2
Semimembranosus	1.9	0.4	0.5	2.2
Semitendinosus	4.3	3.0	1.5	5.1
Biceps femoris	10.9	0.8	0.3	0.6
Sartorius	1.8	1.4	0.1	0.7

Discussion

Except for the adductor longus muscle, all origins were automatically defined with satisfying accuracy compared to previously proposed work [8]. Only limited changes in moment arms would be introduced when applying this automated method. Therefore this method would yield a good initial guess for user evaluation of the muscle attachment points when constructing subject specific musculoskeletal models.

The present work focussed primarily on the muscular anatomy around the hip joint, an area of specific interest in children with CP who often present significant musculoskeletal deformations around the hip joint. Previously proposed methods for subject-specific definition of muscle attachments adapt generic muscle attachments by rescaling and deforming the bone structures [6,8]. this procedure would maximize the use of all available tissue types (fat, muscle and bone tissue) in the MRI data for personalization of generic muscle models. However, future research is needed to estimate the applicability and accuracy of atlas-based non rigid registration for identification of muscle insertion points in subjects with bony deformations at the proximal femur.

References

[1] Sheldon RS, (2004), J. of Biomechanics, 37, 1869–1880
[2] Duda GN, Brand D, Freitag S, Lierse W, Schneider E, (1996), J. of Biomechanics, 29(9), 1185-1190
[3] Arnold A, Salinas S, Asakawa D, Delp S, (2000), Comput. Aided Surg., 5(2), 273-281
[4] Loeckx D, Maes F, Vandermeulen D, Suetens P. (2004), MICCAI , I, 639-646
[5] Maes F, Collignon A, Vandermeulen D, Marchal G, Suetens P. (1997), Trans. Med. Imag., 16(2), 187-198
[6] Arnold A, Blemker S, Delp S, (2001), Annals of Biomedical Engineering, 29, 263–274
[7] Besl P, McKay N, (1992), Trans. Pattern Anal. Machine Intell., 14, 239-256
[8] Kaptein BL, van der Helm FCT, (2003), J. of Biomechanics, 37, 263-273

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DYNAMIC MEASUREMENT OF GASTROCNEMIUS TENDON AND BELLY LENGTH DURING HEEL-TOE AND TOE-WALKING IN NORMALLY DEVELOPING CHILDREN AND ADULTS

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Summary/ Conclusions
During heel-toe walking (HTW) normally-developing (ND) adults and children use isometric contraction of the gastrocnemius muscle belly (GB) to store elastic potential energy in the external gastrocnemius tendon (GT) which is returned during pre-swing. In toe walking (TW) adults have periods of eccentric muscle contraction which is potentially harmful to the muscle. Children, probably due to the higher compliance of their tendons, maintain isometric contraction of the GB throughout TW.

Introduction
Recently the contributions of the GB length to the motion of the knee and ankle has been elaborated during slow walking on a treadmill[1]. Results suggest that the GB contracts isometrically or concentrically during the stance phase of gait and that changes in the length of the GT play a significant role in the total excursion of the musculo-tendinous unit (MTU). Here we use a similar technique to estimate the GB and GT length changes in free walking in adults and children adopting HTW and TW styles.

Statement of clinical significance
Excursion of the GT in HTW and TW protects the GB from damaging eccentric action during the loaded phases of gait, contributes to power generation at the ankle in pre-swing and reduces the motor control burden on the central nervous system. Despite the difference in tendon compliance between adults and children the contributions of the GB and GT to MTU length are similar.

Methods
ND adults (n=6, mean age:32 , range:22-53 years) and children (n=6, mean age:10 , range:7-12 years) were recruited to the study. The Helen Hayes marker set was applied to each subject and a 2D B-mode ultrasound probe was strapped firmly to the shank so that the gastrocnemius musculo-tendinous junction could be clearly seen in the image (see Figure 1). 4 reflective markers were attached to the probe to enable its position to be tracked in space. Each subject was asked to walk at a self selected speed with simultaneous collection of motion tracking data and movie data from the ultrasound machine. Trials were collected for each subject for HTW and TW. GT length was calculated as the distance between the musculotendinous junction and the heel marker. MTU length was calculated from the knee and ankle joint angles using the model described by Eames *et al*.[2]. GB length was calculated as the MTU length – GT length. Lengths were normalised to leg length and normalised length changes were plotted.

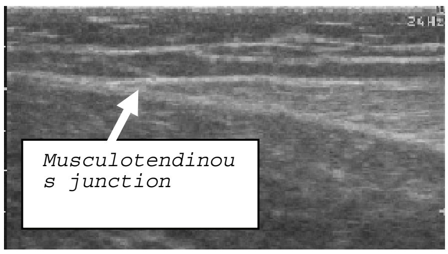

Figure 1: Ultrasound image of musculotendinous junction

Results

The mean normalised length changes of the GB, GT and MTU over the gait cycle are shown in Figure 2 along with the activation patterns of the gastrocnemius muscle.

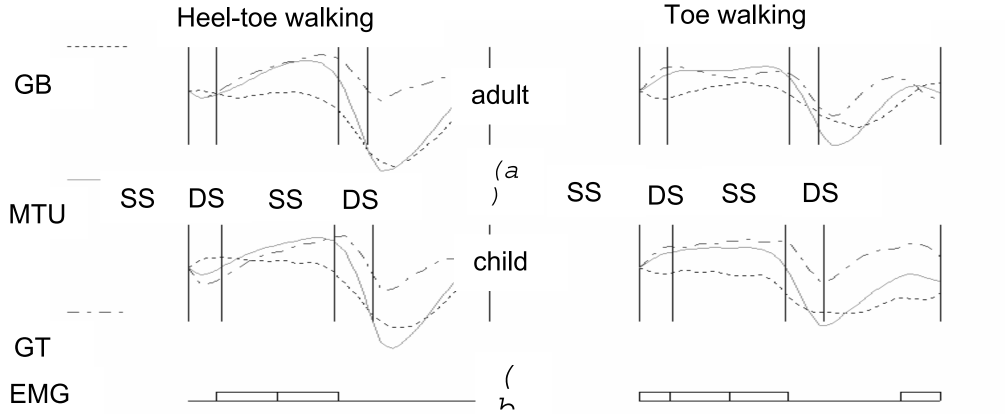

***Figure 2:** (a) Mean changes in normalised length for the MTU, GB and GT in HTW and TW for adults and children (b) typical EMG activation pattern for HTW and TW from initial contact to next initial contact. Periods of single support (SS) and double support (DS) are marked.*

Discussion

In a numerical model of gastrocnemius MTU action during walking, Hof *et al.*[3] suggested that the GB stiffens the MTU so that GT can act to store and release energy at appropriate phases in the gait cycle. Hof went on to predict that the muscle belly would contribute to the shortening of the MTU in pre-swing. Our results validate the work of Hof and agree with the experimental results of Fukunaga[1] for walking at low speeds on a treadmill.

Further, we have shown that voluntary TW alters the interaction between the GB and the GT. Activation of the GB in late swing and loading may act to stiffen the MTU. In children, the GB is able to withstand the large tensile forces acting upon it while the GT stretches. In adults, possibly due to greater body mass and less compliant GT, there are periods of eccentric action carrying the potential for muscle damage.

Our results demonstrate the sympathetic actions of GB and GT during HTW and TW in normally developing subjects, but also suggest a mechanism that explains the deterioration of gait in individuals limited to TW by an underlying condition such as spastic cerebral palsy (SCP). Children with SDCP have GBs of less than half the volume of those of their weight matched ND peers[4]. The weakened GB in SCP may not be able to resist the tensile forces in walking resulting in eccentric muscle action and consequential damage. Progression from TW to crouch may be caused by a requirement to offload the damaged GB.

References

1. Fukunaga T et al. *Proc. R. Soc. Lond. B* (2001) **268** 229-233 In vivo behaviour of human muscle tendon during walking
2. Eames NWA et al *Gait & Posture* (1997) **6** 9-17 Defining gastrocnemius length in ambulant children
3. Hof AL. *J. Biomechanics* (2003) **36** 1031-1038 Muscle mechanics and neuromuscular control
4. Fry et al. *Dev Med Child Neurol* Submitted. Changes to the volume and length of the medial gastrocnemius after surgical recession in children with spastic diplegic cerebral palsy

O-45

ASSESSING THE REGULATORY ACTIVITY OF THE PARETIC LEG IN BALANCE CONTROL

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Summary/conclusions
We measured the regulating activity of the paretic leg during quiet stance and perturbations, using CoP movements and ankle torques. We showed that the contribution of the paretic leg assessed during quiet stance, either based on torques or CoP movements, overestimates the contribution during perturbations, especially in the patients with a clear asymmetric weight distribution.

Introduction
When studying the impact of pathology on balance control, Centre of Pressure (CoP) movements obtained during quiet stance are often used to quantify the regulating activity of the paretic and non-paretic leg. The Centre of Pressure is assumed to reflect the generated ankle torque and as such the generated regulating activity [1]. However, apart of the CoP, also the magnitude and direction of the ground reaction force is an important factor in the calculation of the ankle torque. For pathologies which are characterized by an asymmetry in weight bearing, like stroke, ignoring the ground reaction force might lead to overestimation of the regulating activity. Furthermore, Van der Kooij et al. [2] showed that balance perturbations are necessary in a closed-loop control system in order to distinguish between control activity that truly restores balance and activity which is the consequence of sensory and motor noise.

Statement of clinical significance
For the evaluation of treatments and assessment of the impairments in balance control one needs appropriate descriptors of the regulating activity in the paretic leg. This study compared three different methods to calculate the regulating activity in the paretic leg of stroke patients.

Methods
Nine chronic stroke patients participated in the study. Balance responses were studied during quiet stance and during continuous quasi-random platform movements in forward-backward direction. We recorded body motion and ground reaction forces below each foot to calculate the CoP movements, ankle torques and body sway. The regulating activity was calculated in three different ways. To be able to make a comparison we quantified the paretic regulating activity with respect to the total generated activity. The first two measures were obtained from quiet stance data. First, the Static Balance Contribution based on CoP (SBC_{CoP}) was determined by dividing the root mean square of the CoP velocity ($RMSV_{CoP}$) of the paretic leg by the sum of the $RMSV_{CoP}$ of the paretic and non-paretic leg. Second, the Static Balance Contribution based on ankle torques (SBC_{Torque}) was determined by using the derivative of the ankle torque, calculated from the ground reaction force vector instead of the CoP velocity in the preceding calculation. The third measure was derived from the data of the platform perturbations. The dynamic balance contribution (DBC) [3] of the paretic leg was determined by using a system identification technique to relate the generated ankle torque to the sway movements obtained during quasi-random platform perturbations. By using closed-loop system

identification techniques, only the torques that are used to counteract the sway are used to calculate the regulating activity. Both SBCs and DBC were expressed as fractions and were compared to each other by using a sign test.

Results

The regulating activity of the paretic leg assessed with the aid of platform perturbations (DBC) was significantly ($p<0.05$) smaller as the regulating activity assessed during quiet stance (SBC_{Torque} and SBC_{CoP}), see Figure 1. The difference could not be explained by a difference in weight distribution as the patients did not change their weight distribution while standing on the moving platform. The two SBC measures obtained during quiet stance did not differ significantly from each other. Not all patients showed a marked asymmetry in weight distribution. As we expected that a clear asymmetry in weight distribution could be reflected in a difference between SBC_{CoP} and SBC_{Torque}, we also compared these measures in a subgroup of 5 patients with less then 40% weight on their paretic leg. Indeed, the difference between SBC_{CoP} and SBC_{Torque} was more pronounced in this subgroup as was the difference between DBC and SBC's. (see Figure 1)

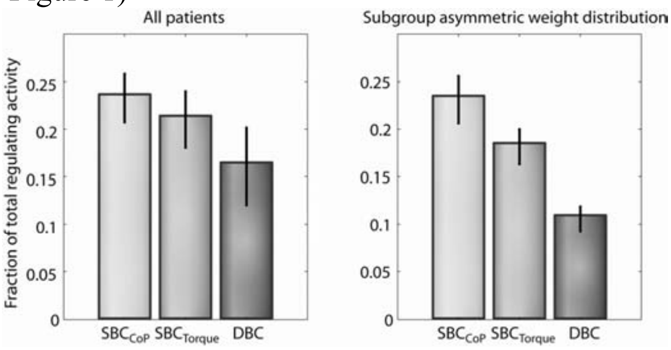

Figure 1. *Mean and standard error of the mean for the three measures of the regulatory activity in the paretic leg in all patients (left) and in the subgroup of patients with asymmetric weight bearing (right).*

Discussion

From a control theoretical point of view, the DBC is the most reliable estimate of the regulating activity. The regulating activity assessed during quiet stance overestimates the contribution of the paretic leg to balance control. One factor could be that the regulating activity assessed during quiet stance is not the same as during balance perturbations, as the last condition is more challenging. The difference between both torque-based measures (SBC_{Torque} and DBC) can also be explained by the relatively larger amount of destabilizing torques, resulting from noise in the system, in the paretic leg compared to the non-paretic leg. These destabilizing torques are part of the calculation of the SBC_{Torque} but not of the DBC.
The estimates of the regulating activity during quiet stance are easier to apply in a clinical setting. In quiet stance, torque based measures should be used especially in pathologies which are characterized by an asymmetric weight distribution. The DBC method is a direct measure of the control activity of each leg, and is more sensitive to differences in regulating activity.

References

[1] De Haart, M., et al., (2004), Arch Phys Med Rehabil, 85, 886-95.
[2] Van der Kooij, H., et al., (2005), Journal of Neuroscience Methods, 145, 175-203.
[3] Van Asseldonk, E.H.F., et al., Experimental Neurology In revision,

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ABNORMAL EMG-ACTIVITY IN PATHOLOGICAL GAIT IN PATIENTS WITHOUT NEUROLOGICAL DISEASES

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Summary/conclusion
The abnormal electromyographic (EMG) activity during gait seen in patients with cerebral palsy (CP) was also found in orthopaedic patients without neurological involvement. The abnormal EMG activity correlated best with muscle weakness. Weakness is a well recognised problem in patients with CP, too. The results of this study suggest that muscle weakness may be more important than spasticity to explain the pathological gait pattern, even in patients with spasticity.

Introduction
Abnormal muscle activity in patients with CP during gait is commonly taken as of spastic origin. Mimicking the individual gait pattern of a given patient with hemiplegic CP, however, produced similar EMG abnormalities in normals [1]. This study investigates the incidence and possible causes of abnormal EMG patterns in patients without neurological diseases.

Statement of clinical significance
In patients without neurological disease, abnormal muscle activity is either an indicator for a compensation strategy or a physiological variety. If the pattern found in CP is found in neurologically normals as well, the question of the origin of this pattern in CP rises. This study contributes to the understanding of gait in CP.

Methods
All patients (n=39) without any neurological disease who were referred to the gait laboratory between January 2003 and March 2005 were included in this study. The primary pathologies varied widely (club feet, ACL-ruptures, Perthes disease, ECF, unclear pain syndromes and more). Since January 2003 the routine of assessment in the gait laboratory was unchanged: the clinical examination of the lower extremities comprised of a test of range of motion (RoM) including all lengths of bi-articular muscles, a manual testing of muscle strength, and an assessment of spasticity (modified Ashworth scale). Instrumented gait analysis was performed (VICON 460, 2 Kistler force plates) including surface EMG of gastrocnemius medialis, tibialis anterior, rectus femoris, and semitendinosus bilaterally. Two data sets were excluded for artefacts. Raw EMG was analysed for abnormal activity. Any EMG activity duration prolonged more than half of maximal activity out of the normal range [2] was considered (figure 1: grey areas):
A) Plantar flexor hyperactivity (gastrocnemius medialis, figure 1a): Too early onset of activation in terminal swing or at latest at initial contact in continuation till foot contact, prolonged activity in stance, usually accompanied by a shut off of the tibialis anterior muscle.
B) Knee extensor hyperactivity (rectus femoris, figure 1b): Activity at mid-stance or later in stance.
C) Hip extensor hyperactivity (semitendinosus, figure 1c): Hamstring activity reaching or exceeding mid-stance.

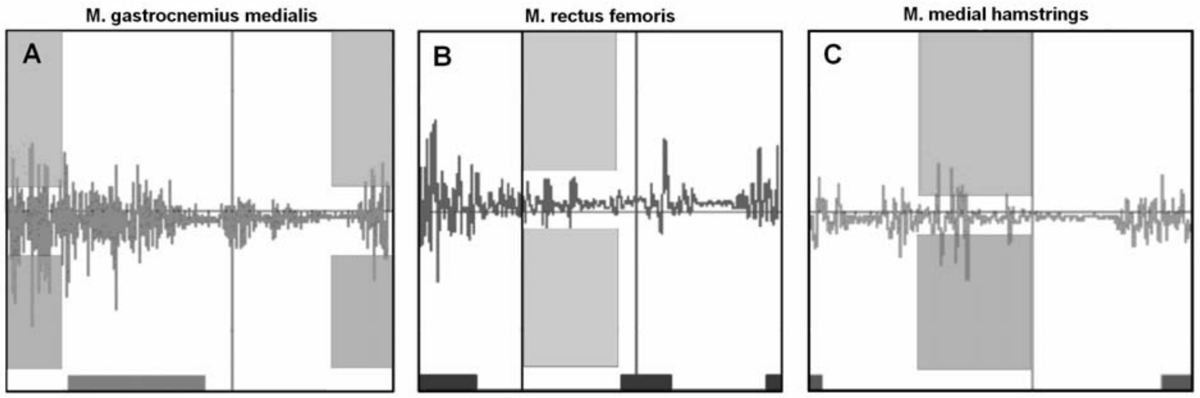

Figure 1. Scoring of pathological raw EMG patterns (figures 1a – c see text)

The 76 extremities were grouped according to the presence of compensations. The number of pathological clinical findings (weakness of antigravity muscles, leg length discrepancy, restrictions in RoM and in muscle lengths) within the 2 groups was compared using the chi-square test. The Spearman rank correlation was used to investigate if variables were interdependent.

Results and Discussion

The presence of abnormal muscle activity correlated best with muscle weakness (Spearman's correlation $p<0.001$). Further correlations were found for reduction of hip extension ($p=0.002$), for reduction of knee flexion with hip extension ($p=0.009$), and for reduced peak plantarflexion. While the first 2 parameters can be explained by rectus femoris shortening, the latter is caused by the overcorrected club feet. Weakness was found in 27 legs at various levels and combinations (plantarflexors 21, knee extensors 9, hip extensors 10) and similarly an abnormal EMG pattern in 47 legs (plantarflexors 26, knee extensors 24, hamstrings 28, co-contractions of hamstrings and knee extensors 13). In the presence of weakness a pathological muscle activity occurred more frequently than without (Chi^2: $p=0.002$), also weak patients without any pathological muscle activity existed. A pathological EMG pattern similar to that in patients with CP was found in neurologically normal patients with a variety of orthopaedic diseases and correlated best with muscle weakness. Patients with CP are known to suffer from weakness underlying their spasticity, too. Weakness, hence, may be the primary problem for the gait disorder, even in neurological conditions. Co-contraction is defined as contraction of agonist and antagonist. If knee extensors and hamstrings are regarded as acting primarily on the knee, the definition is correct. However, if hamstrings are reinforcing weak hip extensors and hence are acting primarily on the hip, the expression "co-contraction" is incorrect, and there would be a good reason why this phenomenon can be seen in patients with normal motor control as well.

References

[1] Romkes J, Brunner R (2005), Proceedings of the 14th Annual ESMAC conference, Barcelona, Spain
[2] Perry J (1992), Gait analysis: Normal and pathological function, Thorofare: Slack Inc., USA.

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AVERAGED EMG PROFILES IN RUNNING COMPARED TO WALKING

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Summary/conclusions
EMGs were collected of 14 muscles with surface electrodes in 10 subjects during walking and running. The EMGs were rectified, interpolated in 100 % of the stride, and averaged over all subjects to give an average profile. Muscles could be divided into a quadriceps, hamstrings, calf and gluteal group, with largely identical profiles within the group. Many muscles show a similar profile in running as in walking. The most notable exception is the calf group, which shows activation in early stance(86-25%), together with quadriceps, instead of in late stance (25-55%) as in walking.

Introduction
In a previous paper [1] a method was presented to quantify the ‘profiles’ of averaged rectified EMGs for human walking in a range of walking speeds. It turned out that the timing of the profiles, when expressed as a fraction of the stride duration, was usually invariable, while their amplitude could vary with speed. The profiles of each muscle could be composed into a limited set of basic patterns, which they had in common within their functional group: calf, quadriceps, hamstrings and gluteal. The aim of the present paper is to make a similar analysis for running.

Statement of clinical significance
When EMG recordings are made in an analysis of pathology in running, these can be compared to the average profiles for healthy young people, as presented here.

Methods
Ten healthy male subjects (age 20.8 ± 1.2 years, mass 71.3 ± 6.3 kg, stature 1.84 ±0.07 m, leg length 0.99 ± 0.05 cm) were included. They walked and ran on a treadmill (ENRAF Entred) at speeds from 1.25 up to 2.25 m.s^{-1} (walking) or 4.5 m.s^{-1} (running). EMGs of fourteen leg muscles of the right leg were recorded. The smoothed (24 Hz) rectified EMGs were linearly interpolated to 100 points per stride, triggered by the right heel contact. Heel contact was obtained from the recording of vertical ground reaction force. At every speed an average over all subjects was obtained[1].

Results
On the basis of their EMG profiles, muscles could be divided in the same functional groups as in walking.
The *vasti* group (vastus medialis and lateralis) showed the same pattern as in walking: one peak from shortly before heel contact (80%) up to 15%. When changing from walking into running at the same speed, there is a 40% increase in amplitude, Figure 1A. At higher running speeds the amplitude increases only little.
The *hamstrings* (biceps femoris, semimembranosus, semitendinosus) showed a pattern with two peaks in late swing and early stance, similar to walking, but some 10% earlier, Figure 1B. The amplitudes of the two peaks changed with speed quite differently for the three muscles.
The muscles of the *calf group* (soleus, gastrocnemius medialis and lateralis, peroneus longus) had a pattern markedly different from walking. In running, activity started shortly before heel contact (86%) and ending before toe-off at 25%, Figure 1C. In walking toe-off is much later (57%) than in running (37%) and peak calf muscle activity is later as well, from 25-55%.
In the *gluteal group* (gluteus maximus and medius) the EMG showed peaks from 88-18% and

from 30-50%, very similar to walking.

Tibialis anterior had continuous activity in swing ending with a peak at 90%, just before heel contact, Figure 1D. In walking the final peak is at heel contact and activity ends at 10% of stance. The start of activity corresponds again with toe-off, but this is much later in walking.

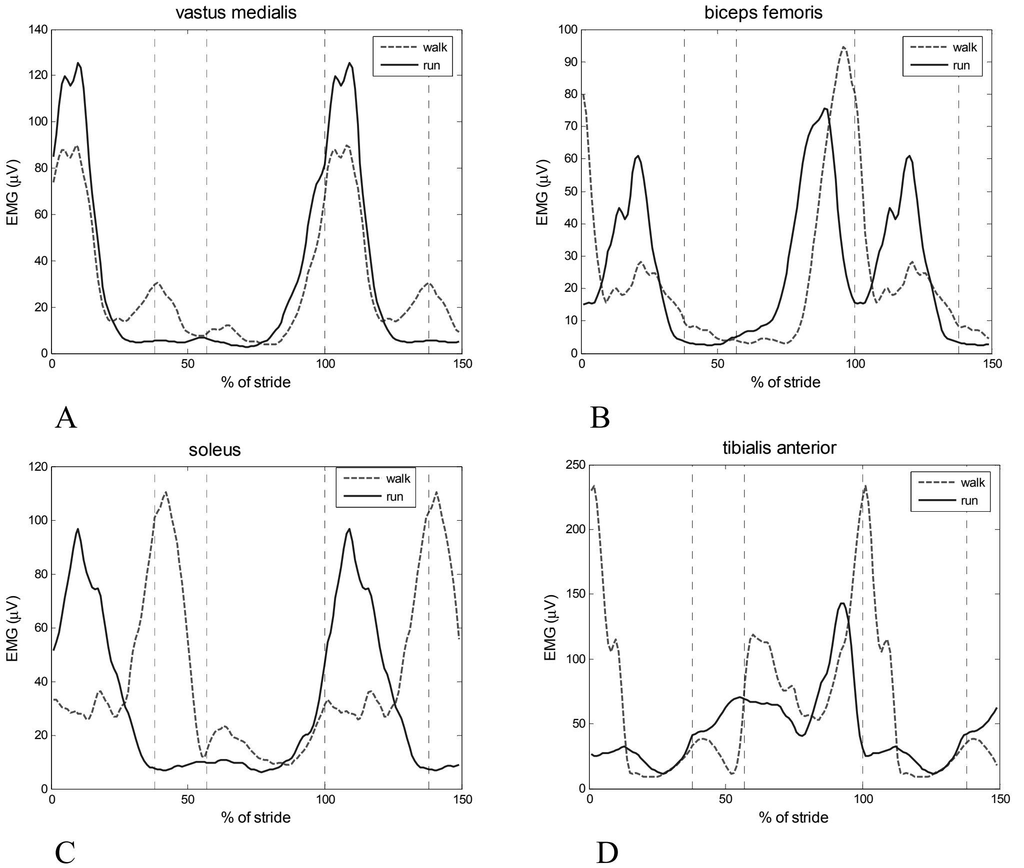

Figure 1 *Average EMG profiles of walking (dashed) and running (solid line) both at 2.25 m/s. A Vastus medialis, B Biceps femoris, C Soleus, D Tibialis anterior. Scale is from 0 – 100 – 50%, to avoid a break at 100%. Vertical dotted lines: toe-off for running (37%) and walking (57%).*

Discussion

The muscles acting on hip and knee show in running largely the same EMG patterns as in walking. Major differences are seen in the muscles around the ankle. The calf group shows an earlier stance activity, well adapted to the much shorter stance phase. This extensor activity is largely simultaneous with massive activity in vasti (knee extensors) and glutei (hip extensors). Tibialis anterior adapts its period of activity to the longer swing phase. The main results are in good agreement with earlier work by Nilsson et al.[2].

References

[1] Hof, A. L. et al. (2002), Gait & Posture, **16**: 78-86

[2] Nilsson, J. et al. (1985), Acta Physiol Scand .,**123**: 457-475

O-48

TRACKING DYNAMIC FOOT PRESSURE PATTERNS IN YOUNG CHILDREN WITH SPASTIC CEREBRAL PALSY

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Summary
A trend of improving foot valgus during the early stages of walking, similar to typical development, is seen in children with cerebral palsy (CP). As they grow, children with CP develop a wide variety of foot pressure patterns and do not follow trends seen in normal development. Most patterns of foot varus or valgus that emerge at age 4 can be predicted from physical exam and gait observational measures at age 2.

Introduction
The natural history of dynamic foot posture is not documented for children with cerebral palsy. Our Gait Laboratory is collecting foot pressure data on a large sample of young children with CP to provide this information. During the first year of the study we reported that pes valgus is present in newly walking children with CP and that this valgus tends to diminish over the first several months of walking[1]. This early trend is similar to that seen in the foot pressure patterns of typically developing children[1]. This ongoing research in the Gait Lab is expected to describe the natural history of foot deformity in children with CP from the onset of walking through adolescence.

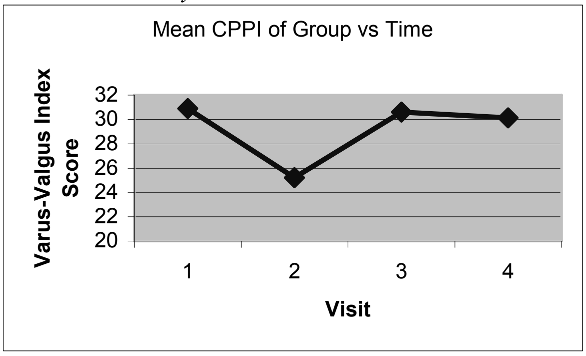
Clinical Significance
Understanding the natural history of dynamic foot posture in children with CP will lead to more appropriate interventions. Predicting which children will develop serious foot deformities allows intervention at an early stage, prevents progression of deformities, and reduces the surgical burden to the child.

Methods
A total of 72 children with CP have participated in our larger study to date. The current analysis includes a subset of 16 children followed every 6 months for two years (+/- 4 months). A brief physical exam was performed and a portion of the GMFM was administered (dimension D) at each visit. Mean age at visit one was 31 months and at visit four was 51 months. Mean GMFM(dimension D) at visit one was 54% and at visit four was 67%. Dynamic foot pressure measurements were collected at each visit using the F-Scan measurement system (Boston, MA). A coronal plane pressure index (CPPI) was utilized to identify dynamic foot valgus or varus pressure patterns during walking[2]. This measurement defines the ratio of medial to lateral pressure impulses in the midfoot and forefoot regions. A trend analysis was performed on the foot pressure data using CPPI across time as the variable of interest. Multiple regression analysis was performed using CPPI at visit four as the outcome variable and a group of 10 physical exam, gross motor, and simple gait measures from visit one as the predictors (forefoot ab/adduction, spasticity, dorsiflexion, hip ER, GMFM, trans-malleolar axis, popliteal angle, foot progression angle in gait, CPPI, and medial midfoot impulse variability).

Results
Figure 1 displays the group mean CPPI at each visit. The initial reduction in valgus seen between visits one and two is significant. The trend of diminishing valgus does not persist across all four visits, as children begin to develop different pressure patterns of varus or valgus with growth.

Results of the multiple regression analysis are presented in tables 1 and 2. Table 1 is a summary which shows that the analysis is significant (Sig F change <.001), the overall correlation of CPPI at visit 4 to the predictor variables (from visit 1) is (R)=.869, with predictive efficacy of 75.5% (R square=.755). Table 2 shows that only 3 of the predictor variables are significant in the regression equation (GMFM, TMA, FPA)

Figure 1: *Trend Analysis: Coronal Plane Pressure Index*



and are necessary in the model to predict CPPI at age 4.

Table 1: *Model Summary of MRA (Predictors = Foot Progression Angle in Gait, dimension D of GMFM, Trans-Malleolar Axis)*

R	R Square	Adjusted R Square	Std Error of the Estimate	R Sq Change	Sig F Change
.869[a]	**.755**	.639	18.5606	.755	**.000**

Table 2: Coefficients

Predictors	B	Std Error	Beta	t	Sig	Partial corr
Forefoot ab/adduction	-.745	.675	-.193	-1.118	.276	-.237
Spasticity (Mod Ashworth)	-1.534	2.913	-.093	-.527	.604	-.114
Dorsiflexion (knee ext)	.192	.615	.051	.312	.758	.068
Hip External Rotation	-.707	.348	-.331	-2.031	.055	-.405
GMFM (dimension D)	-1.394	.476	-.517	-2.929	**.008***	**-.539**
Trans-malleolar axis	-.857	.286	-.465	-3.002	**.007***	**-.548**
Popliteal angle	.099	.304	.060	.326	.748	.071
Coronal Plane Press Indx (1)	.272	.165	.296	1.649	.114	.339
Medial Midfoot Press Variab	.013	.022	.092	.572	.573	.124
Foot Progress Ang in Gait	1.371	.299	.760	4.579	**.000***	**.707**

Discussion

Children with cerebral palsy are predisposed to abnormal foot mechanics for multiple reasons. Impaired tone and motor control disrupt the development of muscular control in the foot. Disturbances in balance and coordination lead to delayed gross motor skill acquisition. Developmental biomechanics of the legs are frequently abnormal in children with CP, producing deforming forces referred to as "lever arm disease" [3]. The purpose of this study was to identify the factors at age 2 that lead to abnormal foot posture at age 4. Our analysis shows that during early stages of walking, the dynamic foot pressure patterns of children with CP follow a trend of diminishing valgus similar to typical development. As children with CP grow and are exposed to abnormal influences, dynamic foot posture changes and a variety of patters emerge. Multiple regression analysis reveals that the primary predictors of foot pressure outcome at age 4 are rotational alignment and gross motor function. Children with a discrepancy between foot progression in gait and trans-malleolar axis on exam, and lower GMFM scores at age 2 developed the most significant valgus foot postures at age 4.

References

1. Lennon N, et.al. Proceedings from the 10th Annual Meeting of GCMAS, 203-204, 2005.
2. Chang CH, et.al., J Pediatr Orthop, 22(6):813-818. 2002.
3. Gage, J, et. al. J Pediatr Orthop B. 10(4):265-74. 2001

O-49

THE INFLUENCE OF FOOTWEAR SOLE HARDNESS ON THE PROBABILITY OF SLIP-INDUCED FALLS IN YOUNG ADULTS

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Summary/conclusions

The results of this study demonstrate that the probability of a slip-induced fall was not influenced by footwear sole hardness. These findings do not support the premise that persons wearing harder soled shoes are at greater risk for a fall once a slip is initiated. It is likely that recovery from a slip event is more related to one's ability to generate an effective response strategy rather than shoe sole hardness.

Introduction

Slips have been recognized as a significant cause of falls and are one of the most common causes of occupational accidents [1]. Several biomechanical factors have been shown to be associated with slip-induced fall events including slip distance and foot velocity during the slip event [2-4]. Once a slip is initiated, prevention of a potentially injurious fall depends on the available friction of shoe/floor interface (i.e., dynamic friction) and the ability of an individual to generate a recovery response [5]. With respect to shoes, harder soled shoes have been shown to provide less dynamic friction than softer soled shoes based on mechanical testing [6]. This would imply that persons wearing harder soled shoes could be at greater risk for falling once a slip has been initiated. Theoretically, a shoe that provides less dynamic friction could result in a greater slip distance and foot velocity, thereby limiting the ability of an individual to generate an effective recovery response. The purpose of this study was to investigate the effects of footwear sole hardness on the probability of slip-induced falls in young adults. It was hypothesized that persons who wear harder soled shoes would demonstrate greater slip distances, slip velocities, and a higher proportion of fall events compared to those who wear softer soled shoes.

Statement of clinical significance

Information provided by this study is important in contributing to our understanding of the factors that may influence slip and fall events and for the design of footwear aimed at reducing slip and fall related injuries.

Methods

Forty healthy young adults between the ages of 22 and 36 years participated in this study. Each was randomized into one of two shoe groups: soft sole hardness (N = 20) and hard sole hardness (N = 20). Two sets of commercially available Oxford style dress shoes (Bates Footwear Inc., Rockford, MI) that differed only in outsole hardness were used in this study. The material and appearance of the uppers of two pairs were identical, and the outsole of each pair was made from Styrene Butadiene Rubber (SBR). One set of shoes had a sole hardness of Shore 75A and was used for the soft shoe group testing, while the other set of shoes had a sole hardness of Shore 54D and was used for the hard shoe group testing. All subjects ambulated at a self-selected fast walking speed across a slippery floor surface consisting of a Teflon panel combined with a soapy water contaminate. The SATRA physical test method (SATRA PM144) revealed that the dynamic frictions for the hard and soft soled shoes on a Teflon surface with soapy water contaminate were 0.09 and 0.16 respectively. A fall arresting harness was used for all walking trials. Slip and fall events were documented using a VICON motion analysis system (120 Hz). In addition, a load cell (Omega Engineering Inc. Stamford, CT) situated between the safety harness and the overhead track was used to measure the forces exerted through the harness. The horizontal displacement of the heel marker at the moment of contact with the floor was used to define the onset of a slip event. Slip termination was defined when the heel marker stopped its forward displacement or the contralateral leg contacted the ground. The determination of a fall was made if the load cell data revealed that the peak force exerted through the harness was greater than 25% of body weight. Slip distance and the peak heel velocity during the slip event were compared between shoe conditions using independent t-tests. Fall probability was compared between shoe conditions using a Chi-square test.

Results

On average, the self-selected fast walking speed of the hard and soft soled shoe groups were similar (Table 1). All subjects in both groups experienced a slip event. Eight of twenty subjects (40%) in the hard soled shoe group experienced a fall, while six of the twenty subjects (30%) in the soft soled shoe group experinced a fall (Table 2). The proportion of fall events in the hard soled shoe group was not statistically different than the proportion of fall events in the soft soled shoe group ($X^2 = 0.440$, $P = 0.51$). In addition, no statistically significant differences were observed between shoe groups for slip distance or average slip velocity (Table 1).

Table 1. Slip characteristics between groups. Mean (SD)

	Hard Shoe Group	Soft Shoe Group	*P* value
Walking velocity (m/sec)	1.9 (0.2)	1.9 (0.2)	0.10
Slip distance (cm)	39.9 (28.7)	40.8 (30.9)	0.92
Slip velocity (cm/sec)	187.5 (98.1)	203.2 (109.4)	0.64

Table 2. Fall proportions between groups

	Fall	Recovery	Total
Hard Shoe Group	8 (40%)	12 (60%)	20
Soft Shoe Group	6 (30%)	14 (70%)	20

Discussion

The results of this study demonstrate that the probability of a slip-induced fall was not influenced by footwear sole hardness. In particular, no difference in the proportion of fall events between two shoe groups was observed. In addition, slip parameters thought to influence fall potential (i.e., slip distance and velocity) did not differ between shoe groups. These findings do not support the premise that persons wearing harder soled shoes are at greater risk for a fall once a slip is initiated. It is likely that recovery from a slip event is more related to one's ability to generate an effective response strategy.

References

[1] Courtney TK, Sorock GS, Manning DP, Collins JW, Holbein-Jenny M. (2001), Ergonomics, 44, 1118-1137.
[2] Brady RA, Pavol MJ, Owings TM, Grabiner MD. (2000), Journal of Biomechanics, 33, 803-808.
[3] You JY, Chou YL, Lin CJ, Su FC. (2001), Clinical Biomechanics, 16, 167-173.
[4] Lockhart TE, Kim S. (2006), Gait & Posture, in press.
[5] Cham R, Redfern MS. (2001), Journal of Biomechanics, 34, 1439-1445.
[6] Redfern MS, Bidanda B. (1994), Ergonomics, 37, 511-524.

O-50

GAIT ANALYSIS IN CHILDREN TREATED NONOPERATIVELY FOR CLUBFOOT: PHYSIOTHERAPY VS PONSETI CASTING

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Summary/Conclusions

There is no significant difference in the likelihood of normal sagittal plane ankle kinematics or normal gait between 2 yr old children with clubfeet treated by either physical therapy (PT) or Ponseti casting. While PT patients are more likely to walk in some degree of equinus, those who have undergone the Ponseti protocol have a higher prevalence of increased stance phase ankle dorsiflexion.

Introduction

Reports of ankle stiffness, loss of ankle power, knee hyperextension, and internal rotation 10 yrs after posteromedial release for clubfeet[1-3] led our institution to support nonoperative treatment. [4] The French PT technique consists of daily manipulation and taping of the infant's foot. [5,6] The Ponseti technique consists of weekly casts to externally rotate the foot about the talar head, usually accompanied by tendoachilles release at 6 wks, with use of an abduction bar nightly for 3 yrs. [7] As we offer both programs, the purpose of this study was to compare gait analysis results in 2 yr olds after successful treatment with PT or Ponseti casting.

Statement of Clinical Significance

PT and Ponseti casting result in similar functional results at 2 yr follow-up. Continued efforts to treat clubfeet nonoperatively are merited.

Methods

Forty-one children with 56 clubfeet treated nonoperatively by Ponseti casting and 47 children with 71 clubfeet treated by the French PT program were enrolled. Pretreatment Dimeglio scores ranged between 10 and 17, where 0 represents normal and 20 a rigid foot. [8] The pretreatment Dimeglio score for both groups was 12.9, indicating similar severity. Mild and moderate feet scoring <10 and clubfeet that had surgery were excluded.

Gait analysis was performed at an ave age of 2.3 yrs (1.9-3.3 yrs). Kinematics were collected at self-selected speed using the VICON 512 motion analysis system. Data was compared to 15 normal 2 yr olds.

Equinus was defined as < 3° DF during stance, calcaneus as PF < one standard deviation(SD) of 2 yr old normal data (< 3° PF at toe off), foot drop as ankle PF > 1 SD of normal (> 9°) in the final 25% of swing, and increased DF > 1 SD of normal (> 13°) during stance phase.

Results

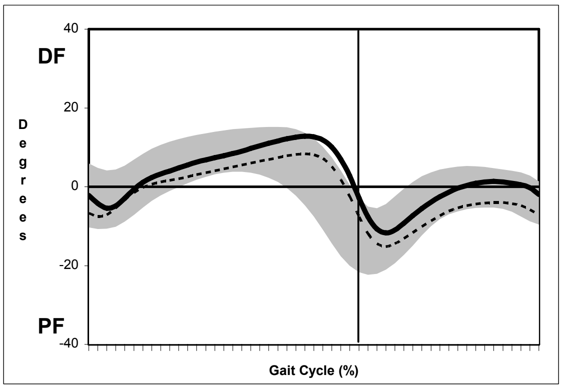
There were no statistically significant differences in cadence, walking speed, or stride time between the treatment groups and normals. There were two clear kinematic patterns identified between groups. More children treated with PT walked in equinus (15%) compared to children who had Ponseti casting (0%)($p<0.05$). Foot drop was present in 18% of PT feet, but only 5% of Ponseti feet ($p<0.05$). Excessive DF was seen more frequently in the Ponseti (43%) compared to the PT group (14%). Overall, ankle sagittal plane kinematics were normal in 62% of PT and 52% of Ponseti feet ($p=0.42$).

Transverse plane kinematics showed a significant difference between both clubfoot groups and normal, with FPA and shank based foot rotation more internal than normal. Clubfeet treated by PT had an internal FPA in 46%, where casted feet had an internal FPA in 21% ($p<0.05$).

Normal gait was present in both 13% of Ponseti and PT feet.

Discussion

Although gait deviations were still seen in this study, results are superior to those of operated clubfeet. [9] PT feet had a tendency toward less DF and increased internal rotation. The sagittal plane findings have led our center and others to adopt early release of the gastrocsoleus. We now anticipate better ankle motion in future patients. Internal rotation was also more frequent in the PT group, likely because the Ponseti group uses a nighttime abduction bar until age 3 yrs. The importance of increased DF seen in some Ponseti feet remains uncertain. Whether this is a predecessor to calcaneus gait, or whether ankle range of motion will diminish with time is unknown.



Sagittal-plane ankle kinematic data over one complete gait cycle for both treatment groups: Ponseti casted (solid) and French physical therapy (dashed). Positive values represent dorsiflexion and negative values represent plantarflexion. Shaded regions represent one standard deviation above and below the average age-matched normal data.

References

1. Karol LA, Concha MC, Johnston CE. (1997), J Pediatr Orthop, Vol 17, pp. 790-5. 2.
2.Deformity and disability from treated clubfoot. Aronson J, Puskarich CL.(1990). J Pediatr Orthop, Vol. 10, pp. 109-119.
3. Davies, TC, Kiefer G, Zernicke RF. (2001), J Pediatr Orthop, Vol. 21: 335-371.
4. Richards BS, Johnston CE, Wilson H. (2005), J Pediatr Orthop, Vol 25, pp. 98-102.
5. Bensahel H, Guillaume A, Czukonyi Z, et al. (1990), J Pediatr Orthop, Vol 10, pp. 189-192.
6. Dimeglio A, Bonnet F, Mazeau P, et al., (1996), J Pediatr Orthop (B), Vol 5, pp. 173-180.
7. Treatment of congenital club foot. Ponseti IV. (1992), J Bone Joint Surg (Am), (1992), Vol.74, pp. 448-454.
8. Dimeglio A, Bensahel H, Souchet P, et al. (1995), J Pediatr Orthop (B), Vol. 4, pp. 129-136.
9. Karol LA, O'Brien SE, Wilson H, et al. (2005), J Pediatr Orthop, Vol 25, pp. 236-240.

O-51

A COMPARISON OF METHODS FOR USING CENTER OF PRESSURE PROGRESSION IN THE CLASSIFICATION OF FOOT DEFORMITY

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Introduction

The pedobarograph, a system that measures plantar pressure, has been successfully applied to patients with neuropathy where the information can be used to prevent pressure-related problems[1,2]. In attempts to utilize the pedobarograph in the evaluation of children with cerebral palsy and associated foot deformities, previous researchers used normalized plantar pressure within different regions of the foot to classify the feet into varus or valgus categories[3,4]. One recent method of pedobarograph analysis used the normal center of pressure progression (COPP) during stance to establish medio-lateral divisions of the foot[5]. The COPP for pathological feet could then be compared to COPP from a normal population. This method required the longitudinal axis of the foot to be subjectively determined by the analyst so the pedobabrograph output could be oriented consistently and comparisons made. The current study expands on that method by introducing simultaneous motion capture to evaluate the difference between subjective and objective methods of long axis determination.

Statement of Clinical Significance

This study reports a new pedobarograph analysis procedure as well as the results for a population of pediatric individuals without gait impairment and compares it to previously published data. This information can then be used in the evaluation and treatment of patients with foot deformity.

Methods

Five successful walking trials were collected for each foot of 23 normal subjects (mean age 11.4 ± 3.3 years, range of 6 to 17 years). One representative trial was selected for each foot for COPP analysis. Because the ultimate goal was to apply this technique to feet with pathology, the custom software developed in our laboratory allows for the analysis of the COPP if the foot shape is atypical and if the entire foot does not make contact with the floor surface. A graphical computer interface allowed the four analysts to specify the long axis of each foot, which was defined as a line connecting a point between the second and third metatarsal and the midpoint of the heel. The medial and lateral borders of the foot were also identified. In addition to this analysis, the 3D kinematic data was used to identify foot landmarks consistent with standard gait analysis, i.e. medial and lateral malleoli, and the midline of the foot at the base of the toes. The center point between the malleoli markers and the toe marker established the long axis of the foot.

The length of the foot was divided into three regions longitudinally (Figure 1). One and two standard deviations from the mean COPP were calculated (Figure 1). These fore/aft and medio-lateral regions were numbered to facilitate classification (Figure 1). The time (% of stance) and location of the COP within the heel, midfoot and forefoot regions were determined so that the COPP could be described temporally as well as in both anteroposterior and mediolateral directions (Figure 2). The rotation angles of the long axis as determined by the two methods was compared, as well as the rotation angles between the four observers.

Results

The average COPP in normal individuals begins in the middle of the heel region at initial contact then progresses forward for 23.8% of stance phase. It proceeds forward through the medial area of the midfoot region for 28.7% of stance. The COPP then moves from the midfoot into the forefoot region where it progresses medially and forward passing just lateral to the hallux. The period of time the COPP spends in the forefoot is 47.5% of stance phase. The results for the comparative analysis of the two methods of long axis determination are shown in figure 2. The A-T method used 3D kinematic data to determine the long axis of the foot. This method is consistent with common gait analysis approaches for measuring foot progression angle. The remaining eight bars in the graph show the long axis angle for the two repetitions of manual long axis selection by each analyst. Three analysts were within one degree of the A-T method, while the fourth analyst, B, was approximately two degrees less than the A-T method. Statistical analysis found significant differences between the methods, but no differences between the analysts. Excellent intra-rater, 0.975, and inter-rater, 0.969, reliability was observed .

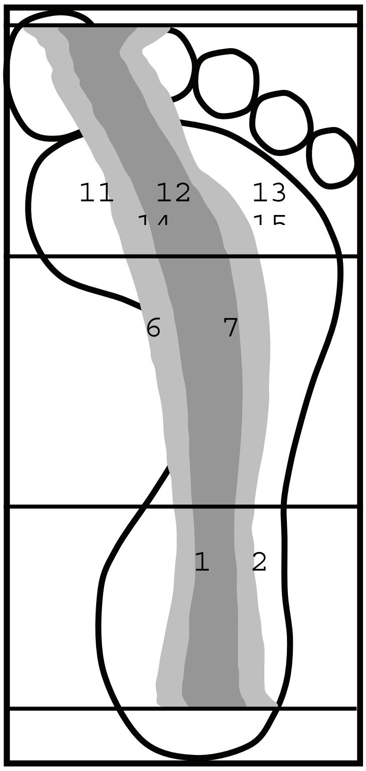

Figure 1. Mean COPP.
Dark gray band is mean +/- one standard deviation; light gray is mean +/- two standard deviations.

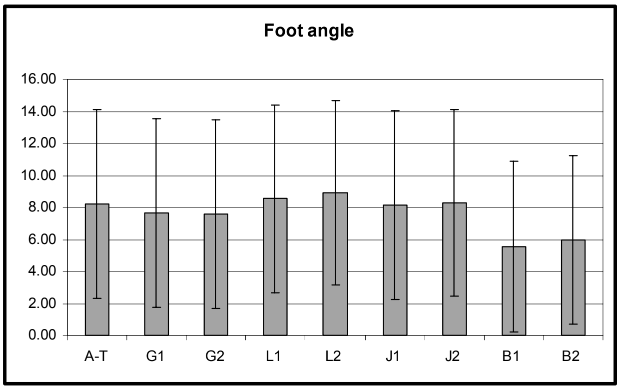

Figure 2. Foot Angle Means and Standard Deviations.
The A-T method employs 3D kinematics to determine the long axis of the foot. The subjective method was performed by four observers (G, L, J, & B), two times for each trial.

Discussion

The statistical differences shown reflect absolute differences less than 2 degrees, these differences may offer little clinical relevance. With the data stratified into the medial to lateral areas of the heel, midfoot and forefoot clinical classifications can be generated, i.e. varus, valgus, severe varus, severe valgus, equinus, calcaneous, and normal. These results give reasonable confidence that manually selecting the long axis is an acceptable method for feet that are normally shaped. However, using the subjective method for malalligned feet or feet in equinus or varus may not reflect the same level of confidence. This technique provides a rational basis for an objective clinical classification of dynamic foot deformities.

References

1. Stokes, IAF, et al. (1975). *Acta Orthop Scand*, 46, 839-47
2. Duckworth, T, et al. (1985). *J. Bone and Joint Surgery*, 67, 79-85
3. Chang, CH, et al. (2002). *J. Pediatric Orthop,* 22, 813-18
4. Bowen, TR, et al (1998). *J. Pediatric Orthop*, 18, 789-93
5. Jameson, GG (2004). *Dev Med Child Neurol*, 99, 34

O-52

BIOMECHANICAL OPTIMIZATION OF ORTHOPEDIC FOOTWEAR FOR DIABETIC PATIENTS USING IN-SHOE PLANTAR PRESSURE MEASUREMENT

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Summary/conclusions
Using in-shoe plantar pressure assessments to evaluate orthopedic footwear can be an effective method to achieve significant pressure reduction at high-risk areas in the diabetic neuropathic foot.

Introduction
Orthopedic footwear is commonly prescribed to diabetic patients with prior plantar foot ulceration. Several studies have reported large inter-individual differences in the pressure reducing effect of various types of custom insoles and shoes [1,2]. Therefore, predicting the pressure reducing effect of footwear remains difficult. As a result, it has been argued that custom footwear should be evaluated and optimized using in-shoe plantar pressure measurement [1,3]. The purpose of this study was to assess the feasibility of using in-shoe plantar pressure measurements to optimize the pressure reducing effect of custom footwear in patients with diabetes.

Statement of clinical significance
Diabetic patients with a prior plantar foot ulcer frequently show recurrence of an ulcer. Although there is limited evidence on the effectiveness of orthopedic footwear in preventing recurrence of plantar ulceration, optimizing the pressure-reducing effects of orthopedic footwear using in-shoe pressure analysis may significantly lower the risk for ulcer recurrence in these patients.

Methods
Ten diabetic patients with peripheral sensory neuropathy and history of plantar foot ulceration who were previously prescribed with orthopedic footwear participated in this study. Using the Pedar-X system (Novel, Germany), in-shoe plantar pressures were measured during four walking trials at a self-selected walking speed. Based on the peak pressure diagrams and values shown on-screen directly after collecting data, a region(s) of interest (ROI) for optimization (i.e. pressure reduction) was defined. This was the region with the highest measured peak pressure and/or with prior ulceration. A maximum of three rounds of footwear modifications were applied. After each round the effect of the modifications on in-shoe plantar pressure was assessed using the same protocol and a standardized walking speed. Footwear modifications included all possible shoe or insole adaptations of which the shoe technician thought they would reduce pressure at the ROI. Criteria for successful optimization were a 25% or more reduction in peak pressure or an absolute reduction of peak pressure below 200 kPa [4]. A detailed analysis of plantar pressure was performed using Novel Multimask software.

Results
A total of 13 ROIs were defined in the 10 patients tested. The number of optimization rounds varied between one and three. Mean peak pressure at the ROI was reduced from 344 (SD 99) kPa in the non-optimized footwear to 229 (SD 73) kPa after footwear optimization. Twelve out of 13 ROIs were successfully optimized by a minimum 25% reduction in peak pressure (mean 33%, range 22% to 50%, see Figure 1 for two examples). In the remaining case peak pressure was reduced below 200 kPa. The maximum time needed for footwear optimization (including

pressure measurement) was 75 minutes.

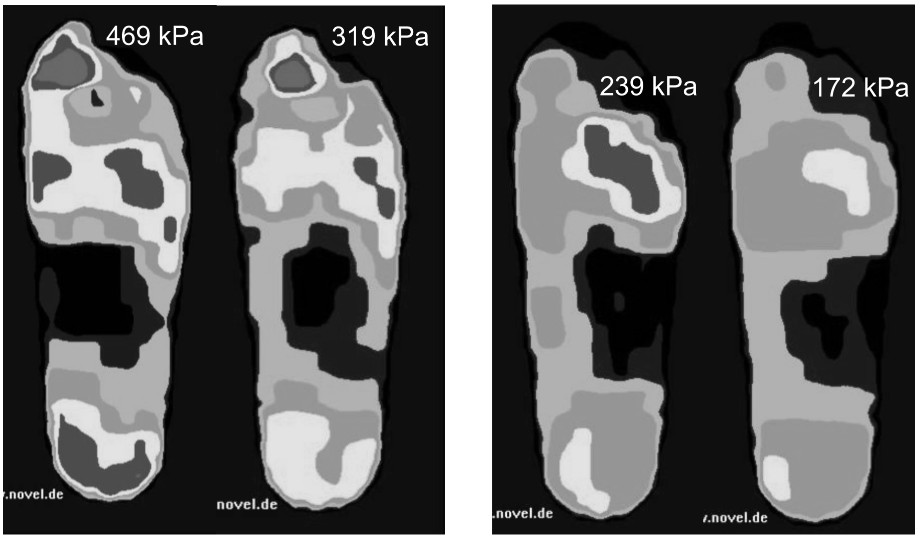

Figure1. Peak pressure diagrams showing the result of footwear modifications made in two cases: a PP reduction from 469 to 319 kPa (=32%) at the hallux in one patient (left two planes) and from 239 to 172 kPa (=28%) at the lateral forefoot in another patient (right two planes)

Discussion

The results showed that the footwear evaluated in this study could be successfully optimized within three rounds of footwear modifications and within a reasonable time frame. These findings suggest that using in-shoe plantar pressure assessments to evaluate orthopedic footwear can be an effective method to achieve significant pressure reduction at high-risk areas in diabetic neuropathic patients. This method provides the clinical team with a more objective approach to footwear prescription and evaluation for the diabetic foot. Whether this approach can contribute to a reduction in plantar ulcer recurrence in these patients remains to be investigated.

References

1. Bus SA, Ulbrecht JS, Cavanagh PR: Pressure relief and load redistribution by custom-made insoles in diabetic patients with neuropathy and foot deformity. *Clinical Biomechanics* 19:629-638, 2004.
2. Guldemond NA, Leffers P, Schaper NC, Sanders AP, Nieman FH, Walenkamp GH: Comparison of foot orthoses made by podiatrists, pedorthists and orthotists regarding plantar pressure reduction in The Netherlands. *BMC Musculoskelet Disord* 6:61, 2005.
3. Cavanagh PR: Therapeutic footwear for people with diabetes. *Diabetes Metab Res Rev* 20 Suppl 1:S51-S55, 2004
4. Cavanagh PR, Ulbrecht JS, Apelqvist J, Stenstrom A, Kalpen A, Bus SA: In-shoe plantar pressure threshold for the prevention of plantar ulcer recurrence. *Diabetes* 51:A255, 2002.

O-53

EFFECT OF SENSORY-THRESHOLD ELECTRICAL NERVE STIMULATION ON MOTOR RECOVERY AND GAIT KINEMATICS AFTER STROKE

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Summary/conclusions
Sensory-threshold electrical stimulation (SES) of the paretic leg in addition to a conventional rehabilitation program was not superior to conventional rehabilitation program alone, in terms of lower extremity motor recovery and gait kinematics of our group of patients with stroke.

Introduction
Sensory input can modulate reorganization of the motor cortex, which may be beneficial in therapeutic interventions to improve motor function in stroke rehabilitation [1]. It has been shown that sub-threshold sensory stimulation of the paretic limb using glove or sock electrodes improved limb function late after stroke [2].

Statement of clinical significance
This prospective randomized controlled trial was designed to assess the effects of SES of the paretic leg on motor recovery and gait kinematics of patients with stroke.

Methods
A total of 30 consecutive inpatients with stroke (mean age of 63.2 years), all within 6 months post-stroke and without volitional ankle dorsiflexion were studied. Both the SES group (n=15) and the placebo group (n=15) participated in a conventional stroke rehabilitation program, 5 days a week for 4 weeks. The conventional program is patient-specific and consists of neurodevelopmental facilitation techniques, physiotherapy, occupational therapy, and speech therapy (if needed). The SES group also received 30 minutes of SES to the paretic leg, 5 days a week for 4 weeks. Stimulation pads were placed at the anatomical localization of the peroneal nerve while the patients were in supine position. Asymmetric biphasic rectangular stimulation at a frequency of 35Hz with a pulse width of 240µs was delivered. The stimulation amplitude was adjusted at each session to the point where the patient perceived a mild tingling sensation (roughly 10mA), but below an observable or palpable muscle contraction. The same set-up was used for the placebo group without any stimulation. Main outcome measures were Brunnstrom's Motor Recovery Stage (BMRS), and time-distance and kinematic characteristics of gait. BMRS I-III indicates more synergistic and mass movements, whereas stages IV-VI indicate isolated and selective movements. Three-dimensional gait data were collected with the Vicon 370 system and processed by the Vicon Clinical Manager (version 3.2) software. Initial and final evaluations were made 1-3 days before and after the 4 weeks of the treatment period. The group means and percentage changes were compared between the SES and the placebo group using non-parametric paired and unpaired t tests. The Chi-square test was used to compare the groups in terms of the number of patients with BMRS for lower extremity I-III or IV-VI.

Results
Age, gender, height, weight, injury and clinical characteristics, time since stroke and walking velocity were all similar between the SES and the placebo group. BMRS improved significantly in both groups ($p<0.05$). In total, 58% of the SES group and 56% of the placebo

group gained voluntary ankle dorsiflexion. Between-group difference of percentage change was not significant (p>0.05). Although walking velocity increased both in the SES (13%) and in the placebo group (13%), the difference between pre- and post-treatment data was not significant. Gait kinematics was improved in both groups but the between-group difference between the groups was not significant.

Table 1. *Outcome measures in the SES group and the placebo group*

Outcome measures	Pre-treatment		*Post-treatment*	
	SES	Placebo	SES	*Placebo*
BMRS lower extremity	3.2±1.6	3.3±1.2	4.1±1.4	*3.5±0.9*
Walking velocity (m/s)	0.31±0.18	0.36±0.22	0.34±0.11	*0.37±0.20*
Step length (m)	0.29±0.11	0.28±0.16	0.32±0.12	*0.30±0.12*
% of stance phase (paretic side)	59.1 ± 3.5	58.1 ± 2.5	59.9 ± 4.7	*58.8±5.7*
Pelvis (°)	11.2±6.7	12.2±3.3	10.0±5.2	*11.0±3.1*
Hip (°)	15.6±9.6	17.3±10.0	16.9±7.8	*17.7±7.9*
Knee (°)	25.4±10.2	27.7±14.9	27.9±11.7	*28.1±9.8*
Ankle (°)	17.4±13.7	16.3±4.6	18.5±5.9	*16.6±9.9*
Maximum ankle DF at swing (°)	-6.2±2.3	-5.9±2.4	-4.5±3.1	*-5.1±1.3*
Maximum ankle PF at initial contact (°)	-1.8±0.9	-3.0±1.5	1.2±7.5	*-2.1±1.2*

Table 2. *Percentage change after treatment in the SES group and the placebo group* p<.05*

Outcome Measures	SES Group (%)	Placebo Group (%)	p-value
ΔBMRS lower extremity	46*	44*	.31
BMRS from (I-III) to (IV-VI)	58	56	.50
ΔWalking velocity (m/sec)	13	13	.97
ΔStep length (m)	18	19	.34
Δ% of stance phase (paretic side)	2	1	.60
ΔPelvis (°)	12	14	.89
ΔHip (°)	4	3	.75
ΔKnee (°)	7	3	.44
ΔAnkle (°)	16	19	.47
ΔMaximum ankle DF at swing (°)	16	14	.64
ΔMaximum ankle PF at initial contact (°)	14	12	.70

Discussion

Yan et al. reported that 15 sessions of motor-threshold functional electrical stimulation, given 30 minutes per session plus standard rehabilitation, 5 days a week, improved lower extremity motor recovery and functional mobility in acute stroke subjects, more than placebo stimulation [3]. In motor stimulation, the current intensity is high enough to exceed motor threshold and evoke muscle contractions which are associated with cutaneous, muscle and joint proprioceptive afferent feedback. However, in sensory stimulation, the low current intensity evokes a sensory reaction without muscle contraction, associated only with cutaneous afferents. Because it has been reported that even the placement of electrodes on the skin is likely to stimulate mechanosensitive nerve fibers, we might have caused an iatrogenic afferent sensory input in our placebo group.

References

[1] Hamdy S. et al. (1998), Nature Neurosci,1,64-8.
[2] Peurala SH. et al. (2002), Clin Rehabil,16,709–16.
[3] Yan T. et al. (2005), Stroke,36,80-5.

O-54

THE IMPACT OF SINGLE EVENT MULTILEVEL SURGERY(SEMLS) FOR SEVERE CROUCH GAIT IN SPASTIC DIPLEGIC CEREBRAL PALSY: OUTCOME AT FIVE YEARS.

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Summary/conclusions
The natural history of gait in spastic diplegia is for deterioration with time[1-4]. The SEMLS programme aimed to correct severe crouch gait and included the wearing of ground reaction ankle foot orthoses (GRAFOs) and physiotherapy. After the intervention, subjects walked with improved extension at the hip and knee, patellar fractures healed, knee pain was relieved and the acute deterioration in functional mobility was reversed. The improvement in severe crouch gait at one year post intervention was maintained at five years, a significant improvement over the natural history. The surgical prescription did not adequately address the involvement of the hip and pelvis and further study into the effect of psoas lengthening at the hip and pelvis and alternatives to excessive hamstring lengthening for the correction of severe crouch gait should be undertaken.

Introduction
Severe crouch gait may be associated with severe anterior knee pain, patellar fractures, fatigue and deterioration in gait pattern[5, 6]. These symptoms may lead the child and family to seek treatment. In crouch gait the antigravity muscles are long/weak (soleus, quadriceps and hip extensors), the hip flexors and hamstrings are usually contracted and the bony levers are often malaligned. We devised a program of reconstructive surgery, which consisted of lengthening the contracted muscle tendon units, realigning the bony levers and supporting the new biomechanical alignment with Ground Reaction AFO's until stable.

Statement of clinical significance
The long-term outcome of SEMLS on severe crouch gait has not been documented. This study is the first to confirm that improvement post-SEMLS at one year can be maintained at five years post-intervention.

Methods
This was a retrospective cohort study, conducted in a tertiary paediatric hospital/ gait analysis laboratory. A consecutive sample of 10 children with spastic diplegic cerebral palsy (GMFCS level II-III) in severe crouch gait, were recruited. Severe crouch gait was defined as knee flexion >30 degrees[6, 7] and ankle dorsiflexion >15 degrees throughout stance.
SEMLS was based on pre-operative gait analysis. Mean of 7 procedures (range 5-10) were undertaken consisting of lengthening of contracted muscle-tendon units and rotational osteotomies and bony stabilization procedures to correct lever arm dysfunction. During the 1st year post SEMLS, GRAFOs were worn routinely and an individually tailored physiotherapy programme in the community provided.
All gait analyses were undertaken barefoot and with usual mobility aids. Post-operative changes were evaluated at one and five years: functional outcome by mobility scales and technical outcomes by 3D kinematics and kinetics, knee radiology and pain scores. Outcomes were analysed with linear regression with robust standard errors.

Results

Mean age preoperatively was 11.4 yr (range 8-14) and at follow-up 17.3 yr (range 13-21). Patellar avulsions and fractures healed (one with fibrous union). Knee pain diminished ($p<0.001$, -3.4, 95% CI –4.8, -2.0).

Improvements in gait that were maintained at five years post SEMLS (Table 1) were: increased knee extension at initial contact, maximum knee extension in stance and knee excursion; achievement of a knee flexor moment in stance; decreased dorsiflexion at initial contact and maximum dorsiflexion in stance; and increased maximum ankle power late stance. Normalised speed and maximum hip extension were unchanged. Mean anterior pelvic tilt increased. Mobility status improved on the Functional Mobility Scale over 500m at 5 years ($p= 0.02$, odds ratio 3.9, 95% CI 1.2, 12.0).

Table 1. Temporal-spatial, kinematic, and kinetic data pre SEMLS, one and five years post SEMLS.

	SEMLS status		
Parameters	**Pre**	**1 year**	**5 year**
	mean ±SD	**mean ±SD**	**mean ±SD**
Normalized Velocity	0.02 ± 0.006	0.02 ± 0.008	0.02 ± 0.008
Mean pelvic tilt	14 ± 12	28 ± 9*	24 ± 9†
Max hip extension ST	17 ± 16	16 ± 12	14 ± 11
Knee extension initial contact	52 ± 7	25 ± 9*	26 ± 10†
Max knee extension ST	44 ± 9	13 ± 9*	17 ± 11†
Knee excursion	21 ± 9	39 ± 9*	37 ± 10†
Max knee flexor moment	0.3 ± 0.2	-0.4 ± 0.3*	-0.2 ± 0.2†
Dorsiflexion initial contact	12 ± 10	3 ± 9*	0 ± 6†
Max dorsiflexion ST	29 ± 9	17 ± 8*	15 ± 6†
Max ankle power generation late ST	1.2 ± 0.6	1.4 ± 0.6	1.8 ± 0.4†

* significance at P value<0.05 between pre and 1 yr post SEMLS
† significance at P value<0.05 between pre and 5 yr post SEMLS
SD standard deviation ST stance phase Max maximum

Discussion

SEMLS for severe crouch gait lead to marked improvements in dynamic knee and ankle function, but pelvic position deteriorated. Hamstrings surgery without psoas lengthening over the brim may be contraindicated. The SEMLS prescription did not involve surgery that would decrease excessive ankle dorsiflexion. The use of GRAFOs in the first year post SEMLS may have contributed to this improvement at the ankle and should continue to be implemented during the rehabilitation period.

References

[1] Bell KJ et al, (2002), Journal of Pediatric Orthopaedics, 22, 677-682.
[2] Gough M et al, (2004), Developmental Medicine and Child Neurology, 46, 91-97.
[3] Johnson DC et al, (1997), Journal of Pediatric Orthopaedics, 17, 392-396.
[4] Wren TAL et al, (2005), Journal of Pediatric Orthopaedics, 25, 79-83.
[5] Miller F, (1995), *Complications in Pediatric Orthopaedic Surgery*, 477-544.
[6] Sutherland DH, Cooper L , (1978), Orthopedic Clinics of North America, 9, 143-153.
[7] Sutherland DH, Davids JR, (1993), Clinical Orthopaedics, 288, 139-147.

O-55

COMPREHENSIVE GAIT ANALYSIS OUTCOMES OF SURGICALLY TREATED IDIOPATHIC TOE-WALKERS

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Summary/conclusions

Idiopathic toe-walkers that develop a gastrocnemius/soleus contracture present with gait deviations in addition to equinus that included mildly increased pelvic tilt, decreased peak knee flexion in swing, and increased external foot progression. Idiopathic toe-walkers had an overall improvement following appropriate surgery, Tendo-Achilles Lengthening (TAL) for those patients with more severe contracture and gastrocnemius lengthening (Vulpius procedure) for those patients with primarily gastrocnemius tightness. Mild remaining deviations included reduced peak dorsiflexion in stance, mildly increased pelvic tilt, and external foot progression.

Introduction

Treatment of idiopathic toe-walking can include surgical lengthening of the gastrocnemius/soleus complex when conservative options have been unsuccessful. Previous outcomes reports of surgery for toe-walking have largely been limited to assessing the sagittal plane motion of dorsiflexion/plantarflexion with minimal quantitative pre-and post-operative analysis [1-3]. The purpose of this study was to evaluate the gait characteristics of idiopathic toe-walkers and to comprehensively assess the outcome of idiopathic toe-walkers treated surgically.

Statement of clinical significance

It is important to understand the full extent of deviations present in idiopathic toe-walkers, so that we may be able to more accurately predict how they will walk following surgical intervention.

Methods

Fourteen children that underwent surgical lengthening for idiopathic toe-walking were included in this retrospective study. 3-D computerized gait analysis was performed on all subjects pre-operatively and at a mean of 13 months (range 10-17) post-operatively. Seven subjects had Vulpius procedures bilaterally (primarily gastrocnemius lengthening), and 7 had TALs (6 percutaneous and 1 open z-lengthening of the Achilles tendon). The mean age at surgery for the Vulpius group was 9.3 years compared to 8.5 years for the TAL group. A TAL was performed when there was limited dorsiflexion with the knee extended (mean -9.1º) and flexed (mean -0.1º). A Vulpius procedure was completed when there was limited dorsiflexion with the knee extended (mean 1.5º) and adequate range with the knee flexed (mean 8.1º). Kinematic and kinetic data for the group as a whole were used to determine pre-operative deviations from normal, changes from pre-to post-operative, and residual deviations post-operative from normal. The idiopathic toe-walkers were also grouped by surgical intervention (TAL vs. Vulpius) for comparison.

Results

Compared to normative controls, idiopathic toe-walkers as a group had significantly greater anterior pelvic tilt, decreased peak knee flexion in swing, and greater external foot progression, in addition to the expected increased ankle plantarflexion pre-operatively (Table 1). When grouped by surgical type, the TAL group walked pre-operatively with significantly less peak stance phase dorsiflexion (mean -7.8º) than the Vulpius group (mean 6.1º). Kinetically, the subjects had lower peak ankle power generation in late stance prior to surgery (regardless of surgery group). Post-operatively, peak knee flexion normalized and anterior pelvic tilt decreased by a mean of 4.4 degrees for the TAL Group only ($p<0.01$). The foot progress angle did not change pre- to post-operative remaining significantly more external than normal. Both the TAL and Vulpius groups improved significantly in dorsiflexion in stance and swing to equivalent values. Post-operatively, both surgery groups had significantly improved ankle dorsiflexion measures on physical exam (knee extended: TAL = 5.5º, Vulpius = 6.6º; knee flexed: TAL = 16.3º, Vulpius = 15.3º). Overall, stride length increased and cadence decreased leading to equivalent velocity pre- to post-operative. The TAL and Vulpius groups had equivalent peak ankle power generation pre- and post-operative, and both normalized after surgery. Post-operatively, idiopathic toe-walkers continued to exhibit deviations significantly different than normal subjects with greater mean pelvic tilt, greater mean external hip rotation in stance, lower peak ankle dorsiflexion in stance and swing, and greater external foot progress angle (Table 1).

*Table 1. Idiopathic Toe-Walker (ITW) gait data pre and post gastrocnemius/soleus lengthenings along with lab-based normal data. *Significance at $p<0.01$.*

	Pre ITW	Post ITW	*Lab-Based Normals*	pre- to post-op p-value	pre-op to normal p-value	post-op to normal p-value
Mean Pelvic Tilt	17.4 ‡	15.1	*11.4*	0.005*	0.000*	0.001*
Mean Hip Rotation, Stance†	-12.6	-11.1	*-5.3*	0.447	0.001*	0.010*
Peak Knee Flexion, Swing	51.6	55.9	*56.2*	0.002*	0.000*	0.770
Dorsiflexion, Initial Contact	-18.6 ‡	-7.7	*-2.5*	0.000*	0.000*	0.000*
Peak Dorsiflexion, Stance	-0.9 ‡	9.3	*13.9*	0.000*	0.000*	0.000*
Peak Dorsiflexion, Swing	-13.3 ‡	-1.6	*2.8*	0.000*	0.000*	0.000*
Mean Foot Progression, Stance†	-12.1	-12.9	*-7.4*	0.464	0.004*	0.000*
Cadence(steps/s)	1.146	1.080	*0.996*	0.002*	0.000*	0.001*
Stride Length (cm)	99.2	109.3	*110.9*	0.000*	0.000*	0.618
Speed (cm/s)	112.5	116.8	*109.8*	0.092	0.353	0.017

† Negative values are external angles. ‡ When analyzed by grouping into tendo-Achilles lengthening and Vulpius surgeries, they were different pre-operative, but equivalent post-operative.

Discussion

Following surgical lengthening for gastrocnemius/soleus contracture, idiopathic toe-walkers had an overall improvement in gait parameters and function, but exhibited mild residual deviations. Increased external foot progression seemed to be accounted for by pre-existing external rotation of the femur (9º more on physical exam) and/or tibia (7º more on physical exam). This study did not demonstrate superiority or equivalence between the two surgical interventions of a TAL and Vulpius procedures for idiopathic toe-walkers. Post-operatively, the groups had an overall equivalent walking pattern, but were not equivalent pre-operatively. The TAL surgery did not produce harmful short term results in these non-neurologically involved patients such crouch gait. These results should not be applied to patients with cerebral palsy. The TAL and Vulpius surgery groups of idiopathic toe-walkers had mean post-operative outcomes that were largely within the standard deviation range of normal, indicating that both surgical procedures are effective when applied to appropriate patients.

References:

1.Kogan M et al. (2001), *J Pediatr Orthop*, 21, 790-791.
2. Stott NS et al. (2004), *J Pediatr Orthop*, 24, 63-69.
3. Stricker SJ et al. (1998), *J Pediatr Orthop*, 18, 289-293.

O-56

THE EFFECT OF INCLUDING S2 ROOTLETS IN SELECTIVE DORSAL RHIZOTOMY SURGERY

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Summary and Conclusions
One and two year outcomes for selective dorsal rhizotomy surgery spanning L1-S1 and L1-S2 rootlets were essentially equivalent.

Introduction
Selective dorsal rhizotomy (SDR) has been used to reduce tone and increase function in patients with cerebral palsy (CP). Surgical techniques vary, but the typical method involves micro-dissection and electrophysiological testing. One element of the technique that has remained a topic of debate is whether S2 level rootlets should be included. In a study of 85 subjects, Lang found that sparing S2 rootlets leaves "*functionally impairing spasticity*" in the plantarflexors [1]. Lang's study did not include quantitative gait measures as part of the outcome. Conversely, Molenaers' study of 12 subjects suggested that inclusion of S2 rootlets, while producing 1-year outcomes equivalent to the S1 surgery, lead to loss of pelvic tilt, hip extension and knee extension improvements 2 years post-SDR [2]. Molenaers' study did not report plantarflexor spasticity outcomes.

Statement of Clinical Significance
It is important to know whether or not S2 rootlets should be included in SDR surgery.

Methods
Following ethical approval subjects were retrospectively identified as follows: *i*) gait analysis 0-18 months before SDR (*pre*), 8-36 months after SDR (*post #1*), and 6-24 months after *post #1* (*post #2*), *ii*) SDR at Gillette Children's Specialty Healthcare or Shriner's Hospital for Children–Twin Cities. Other clinical patient criteria and surgical details found in prior publications [3]. Groups were created based on whether S2 rootlets had been included (S2) or not (S1). A linear mixed model analysis was used to assess kinematic outcome measures over three time points (*pre*, *post #1*, and *post #2*), while plantarflexor spasticity outcome *pre→post #1* was assessed using repeated measures ANOVA (SPSS 13.0.1, SPSS, Inc., Chicago, USA).

Results
There were 97 subjects with *pre* and initial follow-up (*post #1*) data and 27 subjects with subsequent follow-up data (*post #2*) [Table 1]. Many subjects underwent orthopaedic surgery following *post #1*, leading to the significant "drop out" rate.

All kinematic measures improved *pre→post #1* and were unchanged from *post #1→post #2,* except mean pelvic tilt, which worsened for both groups and both intervals. No differences were found in the response of kinematic variables based on level of SDR (*i.e.* no pre/pst by S1/S2 interactions); in fact the smallest p-value for an S1 vs. S2 interaction was p = 0.51.

Spasticity, as measured by Ashworth score, was reduced equally and significantly for both groups during the *pre→post #1* interval (S1: 3.1→1.7, S2: 3.0→1.6).

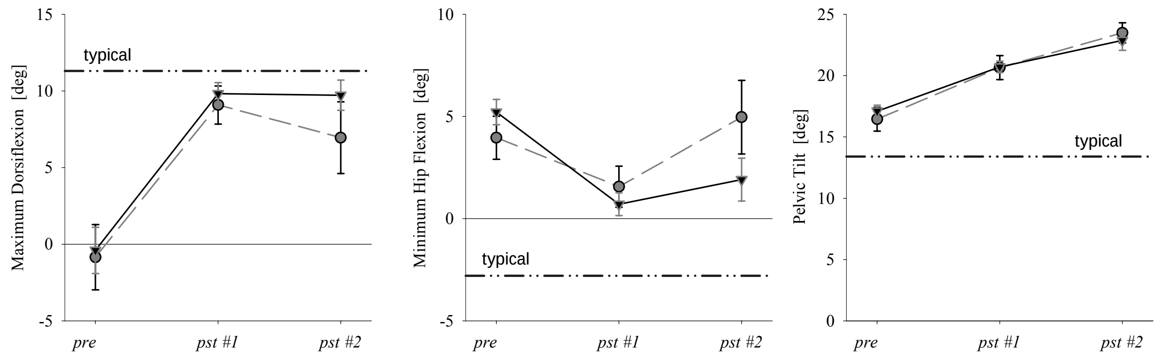
Table 1. Subject demographics

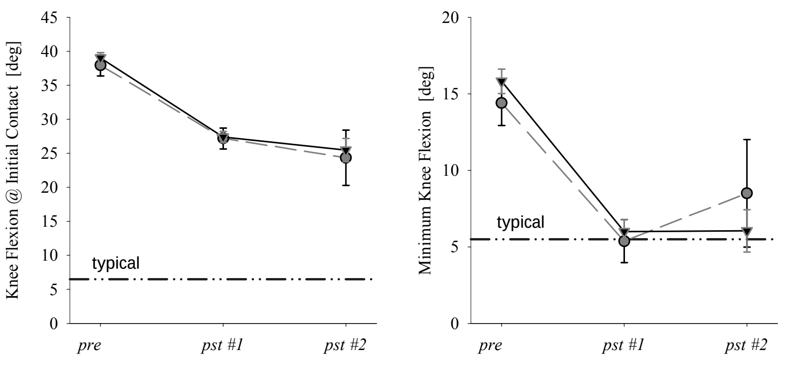
Group	Age pre	N (*post #1*)	Follow-up time (*post #1*)	N (*post #2*)	Follow-up time (*post #2*)
S1	6.8 (2.3)	28	1.1 (0.3)	7	3.6 (0.6)
S2	5.5 (1.3)	69	1.2 (0.7)	20	3.5 (0.5)

Key: mean (standard deviation), all times in years.

Discussion

This study showed that response of subjects to L1-S1 and L1-S2 SDR surgery was equivalent for a specific set of kinematic and spasticity outcome measures. There has been significant controversy over whether S2 rootlets should be included in SDR surgery. Supporting S2 inclusion was Lang's study showing that sparing S2 left residual plantarflexor spasticity. Opposing S2 inclusion was Molenaers' study showing that including S2 promoted crouch gait. This study appears to contradict *both* of those prior studies. The data analyzed here shows equivalent outcomes, between S1 and S2 surgeries, for both plantarflexor spasticity and selected kinematic variables. The explanation for this seems to lie in clinical principles underlying the surgical technique as applied here. For the majority of subjects, rootlets demonstrating pathological electrophysiology were sectioned while rootlets with appropriate response were spared (*n.b.* some subjects did have S2 rootlets spared for a variety of reasons not directly related to electrophysiological response). Clearly the question of including vs. sparing S2 rootlets remains unresolved. Further analysis into long(er) term outcomes, including foot-related outcomes, is warranted.





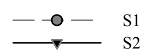


Figure 1. Mean and 95% confidence interval for kinematic outcome measures are shown. No significant interactions were found. The *trends* that appear (minimum hip flexion and knee flexion and maximum dorsiflexion) favor the S2 group. Disconcertingly, mean pelvic tilt showed a deteriorating trend in both groups and both intervals.

References

1.Lang FF *et al*., Neurosurgery, **34**:847-853

2.Molenaers G, *et al*., 13th ESMAC, Warsaw, 2004.

3. Schwartz *et al*. 14th ESMAC, Barcelona, 2005

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DYNAMIC FUNCTION AFTER ANTERIOR CRUCIATE LIGAMENT RECONSTRUCTION IS RELATED TO GRAFT CHOICE

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Summary/Conclusions
In this study clear differences were observed in the knee kinematic and kinetic profiles between patients who had undergone hamstring or patellar tendon anterior cruciate ligament (ACL) reconstruction. Patients with patellar tendon grafts demonstrated reduced external knee flexion moments whereas patients with hamstring grafts demonstrated reduced external knee extension moments. These results suggest that there are graft-specific differences in knee biomechanics after ACL reconstruction that appear to relate to the donor site.

Introduction
ACL reconstruction is a common procedure. However, debate continues as to whether the patellar tendon (PT) or the hamstring tendon (HS) graft is preferable. A number of studies have examined the effect of ACL reconstruction on joint kinematic and kinetic patterns during gait. These have shown that although there is a tendency towards gait normalization after the reconstructive procedure, altered moments about the knee in the flexion-extension axis still persist [1-4]. It has been suggested that abnormal gait pattens might have implications for the long term development of pathologic knee conditions [4, 5]. However, in the majority of these studies, only patients who had received a PT graft were examined. Therefore, the purpose of this study was to investigate biomechanical differences associated with the use of both HS and PT grafts during walking and also during higher impact activities that involved single limb landing.

Statement of Clinical Significance
As both hamstring and patellar tendon grafts are used for ACL reconstruction, it is clinically relevant to understand the biomechanical differences in knee function associated with each graft. The reduced knee flexion angles and moments we observed in the patellar tendon patients may place this group at greater risk of developing degenerative joint changes.

Methods
Two experiments were conducted. In experiment 1, the gait patterns of 17 PT patients, 17 HS patients and 17 controls matched for activity were compared. In experiment 2, single-limb landing patterns for vertical and horizontal hops were compared between 17 PT patients, 17 HS and 12 control subjects. ACL patients were tested 9 to 12 months following surgery. For both experiments a 3-dimensional motion analysis and force plate system was used to determine sagittal plane kinematics and kinetics of the lower limb.

Results
In experiment 1 (walking) there were significant differences in the moments about the operated knee that related to graft type. The PT patients had a reduced external knee flexion moment at mid stance whilst the HS patients had a reduced external extension moment at terminal stance. Experiment 2 (landing) results for both the horizontal and vertical hops also showed a reduction in the external flexion moment about the operated knee for the PT group compared to the non-operated side, the control subjects and the operated side of the HS patients, but no differences compared with the operated side of the HS group. In the PT group the maximum knee flexion angle in the operated limb was also reduced compared to the non operated limb for both horizontal and vertical hops. During landing from a horizontal hop the external knee flexor moment was the predominant moment for the control group but not for patient groups. For the vertical hop the ankle dorsiflexion moment predominated in all subject groups. The patient

groups also appeared to increase their hip flexion moment to compensate for reductions about the knee.

Discussion

This study presents the first attempt to compare knee biomechanics during walking and landing between the two most commonly used graft types for ACL reconstruction and an activity matched control group. The most notable difference between the groups was seen in the moments about the knee. The patellar tendon group had a reduced external knee flexion moment at mid stance in gait and during single limb landing. The hamstring group had a reduced external extension moment at terminal stance. As the principle difference between the groups appeared to be the graft itself, these biomechanical changes may result from the effects of graft harvest. There is some evidence to suggest that reduced knee flexion interferes with the normal ability of the knee to absorb shock [6], and may lead to early degenerative joint changes. If this is the case, the patellar tendon patients would be expected to be at greater risk of this. Clearly further long term studies are required to explore the relationship between knee biomechanics and osteoarthritis. This is an area worthy of future research as both graft types are currently used for ACL reconstruction to allow patients to return to sporting activities.

References

1. Bush-Joseph, C.A., et al., *Dynamic function after anterior cruciate ligament reconstruction with autologous patellar tendon.* American Journal of Sports Medicine, 2001. **29**(1): p. 36-41.
2. Devita, P., T. Hortobagyi, and J. Barrier, *Gait biomechanics are not normal after anterior cruciate ligament reconstruction and accelerated rehabilitation.* Med Sci Sports Exerc, 1998. **30**(10): p. 1481-8.
3. Ferber, R., et al., *Gait mechanics in chronic ACL deficiency and subsequent repair.* Clin Biomech, 2002. **17**(4): p. 274-85.
4. Georgoulis, A., et al., *Three-dimensional tibiofemoral kinematics of the anterior cruciate ligament-deficient and reconstructed knee during walking.* Am J Sports Med, 2003. **31**(1): p. 75-9.
5. Noyes, F.R., et al., *The anterior cruciate ligament-deficient knee with varus alignment. An analysis of gait adaptations and dynamic joint loadings.* Am J Sports Med, 1992. **20**(6): p. 707-16.
6. Cook, T., et al., *Effects of restricted knee flexion and walking speed on the vertical ground reaction force during gait.* J Orthop Sports Phys Ther., 1997. **25**(4): p. 236-44.

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RECOVERY OF MUSCLE STRENGTH FOLLOWING MULTI-LEVEL ORTHOPAEDIC SURGERY IN DIPLEGIC CEREBRAL PALSY

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Summary/conclusions

This study was undertaken to establish the length of muscle strength recovery following multi-level orthopaedic surgery in diplegic cerebral palsy. We demonstrated that the majority of lower limb muscle groups are weaker than pre-operatively at 1 year from operation, despite intensive physiotherapy.

Introduction

We have previously shown that children with cerebral palsy are weak and at 6 months post multi-level surgery there is a significant reduction in both muscle strength and motor function, despite significant improvements in gait at this stage. Following intensive physiotherapy focusing on muscle strengthening at 6 months post-op, partial recovery of muscle strength was observed [1]. The aim of the present study was to assess the continuing recovery of muscle strength by following-up patients to 1 year post-surgery.

Statement of Clinical Significance

Surgical techniques and post-operative physiotherapy regimes following multi-level surgery should improve further in order to preserve and improve muscle strength.

Methods

Twenty children with spastic diplegia who underwent single-stage multi-level surgery (mean age at surgery 12.5 years, GMFCS level I-III) were matched for age, sex and functional level and randomised to two groups. Both groups immediately commenced routine post-operative physiotherapy, and at 6 months post-op additional strength training regimes were administered three times weekly for six weeks. Group A (n=11) undertook a progressive resistance training regime using weights and Group B (n=9) an active exercise regime against gravity only. A control group of 10 children with cerebral palsy/spastic diplegia (GMFCS levels I-II, mean age 11 years) were also recruited for comparison.
Gait (Vicon 612 system), isometric muscle strength (hip flexors, hip extensors, hip abductors, knee flexors, and knee extensors at 90° and 30° of flexion, MIE digital dynamometer) and motor function (GMFM-88) were assessed in both groups pre-operatively, at 6 months post-operatively prior to strength training, and immediately after strength training. The same measurements were repeated at 12 months. Within group comparisons were analysed using paired sample t-tests. Differences between the two groups were evaluated using analysis of co-variance. P=0.05 was considered the level of significance.

Results

In the control group, no changes in muscle strength were observed at 6 or 12 months. The GMFM showed trends towards deterioration without reaching statistical significance. Sagittal kinematics and some kinetic parameters showed significant deterioration at 1 year post-baseline. In both surgical groups, significant decrease of strength was observed post-surgery and partial improvement following physiotherapy at 6 months. Strength improved further at 1 year post-op but did not reach pre-op values, with the exception of hip abductors and knee extensors at 30° (Table 1). No significant difference was observed between the two surgical groups undergoing different physiotherapy regimes.

*Table 1. Muscle strength measurements pre-operatively and at 6 months (before pre- and post-strengthening physiotherapy) and 1 year post-operatively. *Lack of statistical significance indicates return of muscle strength to pre-op values*

Muscle group Nm/kg	**Pre – op mean (SD)**	**6 m I Pre-physio mean (SD)**	**6 m II Post-physio mean (SD)**	**1 year mean (SD)**	**p_i pre-op/ 6 m I**	**p_{ii} 6mI/ 6mII**	**p_{iii} pre-op/ 1 yr**
Hip flexors	0.5(0.2)	0.3(0.1)	0.4(0.2)	0.4(0.2)	0.000	0.000	0.008
Hip extensors	2.5(0.9)	2.0(0.8)	2.2(0.7)	2.2(0.7)	0.000	0.021	0.002
Hip abductors	0.6(0.2)	0.5(0.2)	0.6(0.3)	0.7(0.3)	0.002	0.000	**0.296***
Knee flexors	1.0(0.3)	0.4(0.2)	0.5(0.2)	0.5(0.2)	0.000	0.000	0.000
Knee ext 90°	1.6(0.7)	1.1(0.4)	1.3(0.5)	1.3(0.5)	0.000	0.000	0.001
Knee ext 30°	0.6(0.4)	0.3(0.2)	0.5(0.2)	0.5(0.2)	0.008	0.000	**0.077***

Both surgical groups showed significant improvement in gait parameters at 6 months, (with Group A (progressive resistance training) showing some advantages over Group B. Some deterioration was observed at 1 year but patients maintained the benefits of the operation, when compared to pre-op values (see Table 2). The GMFM deteriorated at 6 months and returned to pre-op levels at 1 year.

*Table 2. Kinematic parameters pre-operatively and at 6 months (before pre- and post-strengthening physiotherapy) and 1 year post-operatively. *Lack of statistical significance indicates return of muscle strength to pre-op values*

Kinematic parameters(°)	**Pre – op mean (SD)**	**6 m I Pre-physio mean (SD)**	**6 m II Post-physio mean (SD)**	**1 year mean (SD)**	**p_i pre-op/ 6 m I**	**p_{ii} 6mI/ 6mII**	**p_{iii} pre-op/ 1 yr**
Range pelvic tilt	10.2(3.0)	8.1(2.7)	7.9(3.3)	7.7(3.4)	0.000	0.714*	0.000
Range knee flexn	31.9(12.9)	49.0(16.1)	47.8(11.7)	43.5(14.1)	0.000	0.004	0.000
Range ankle d.flex	31.0(13.9)	11.8(7.5)	12.6(8.1)	21.1(9.1)	0.000	0.268*	0.000
Max knee extn	32.9(24.7)	9.3(17.4)	10.5(15.5)	16.2(16.7)	0.000	0.009	0.000
Knee flexion IC	44.1(19.7)	21.1(14.1)	21.1(13.7)	28.4(11.6)	0.000	0.362*	0.000
Max hip IR	22.7(12.6)	4.8(17.6)	8.8(10.1)	11.4(9.4)	0.000	0.190*	0.000

Discussion

We have previously demonstrated significant reduction of muscle strength in diplegic patients at 6 months following multi-level surgery and improvement in their kinematic parameters. We are now presenting the 1-year follow-up of these patients. Despite muscle strength improvement following intensive physiotherapy focusing on strengthening at 6 months post-op, no significant further improvement was observed and most muscle groups remained weaker than pre-operatively. The hamstrings appeared to be most affected by surgery. The increase in strength of the hip abductors could be explained by the lever-arm correction achieved through derotation femoral osteotomy. Despite the loss in muscle strength, gait parameters improved significantly through surgery, while these deteriorated in the control group.

References

[1] Thompson et al, Gait & Posture 22S(2005)S1 - S53; S2-S3

O-59

AUDITORY-PACED WALKING FOLLOWING STROKE

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Summary/conclusions
Gait coordination is usually compromised following stroke. Auditory pacing seems to be an expedient means to improve hemiplegic gait coordination, although, to date, detailed studies of the temporal coupling between pacing signal and gait characteristics have been lacking. In the present study this coupling was studied in both healthy and hemiplegic participants walking on a treadmill. Hemiplegics predominantly coordinated the movements of their non-paretic lower limb to ipsilateral auditory pacing stimuli. This instance of perceptual-motor anchoring may provide promising possibilities for future therapeutic interventions.

Introduction

Stroke patients' gait coordination is often hampered by a marked interlimb asymmetry, impaired timing of thoracic-pelvic rotations, and altered coordinative variability [1-2]. The use of auditory pacing as a therapeutic technique may positively affect hemiplegic walking, e.g., in terms of walking velocity, stride length and gait symmetry [2]. Unfortunately, the temporal coupling between pacing signal and gait has not been studied thus far, in spite of the fact that a detailed analysis of auditory-motor coordination might provide valuable information about the underlying motor control processes. The purpose of this study was to examine auditory-motor coordination in paced walking following stroke, in particular the coordination between footfalls and the auditory pacing signal.

Statement of clinical significance

A better understanding of auditory-paced walking in both unimpaired and pathological gait coordination might lead to both evidence- and theory-based rehabilitation goals, cf. [2].

Methods

Six individuals with chronic post-stroke hemiparesis and six age-matched healthy controls volunteered in the study. First, the stride frequency was determined when participants walked at their comfortable velocity on a treadmill. Subsequently, while keeping velocity constant, stride frequency was manipulated by means of auditory pacing. The pacing frequency increased in three steps from 90%, via 100%, to 110% of the stride frequency observed in unpaced comfortable treadmill walking. Participants were instructed to synchronize heel strikes with ipsilateral auditory pacing stimuli, that is, heel strikes of the left (right) foot had to be synchronized with auditory beeps presented at the left (right) ear. Auditory-motor coordination (i.e., mean absolute error between pacing frequency and stride frequency $|\Delta f|$ and variability of the relative phase between ipsilateral instants of auditory pacing and heel strikes σ) was compared between groups.

Results

All control participants and four stroke patients coupled their stride frequency to the prescribed auditory pacing frequency. On average $|\Delta f|$ was 0.05 strides/min. In Figure 1, individual spectrograms are shown. The two patients with poor frequency coupling made more strides in the 90% pacing condition and fewer strides in the 110% pacing condition than prescribed (Figure 1, lower panel). Note that stroke patients' spectrograms exhibited larger stride-frequency variability as well. This tendency was statistically significant at the group level, i.e., stride interval variability was larger ($p < .001$) for the hemiplegics (53ms) than for the control participants (15ms).

For trials with clear frequency coupling ($|\Delta f| < 0.05$), the variability of the relative phase between ipsilateral instants of auditory pacing and heel strikes σ for the left and right side were compared. In the control group variability of relative phase did not differ significantly between the left (7.3°) and right (7.1°) side of the body, whereas in the hemiplegic group σ was significantly ($p < .05$)lower for the non-paretic side (9.5°) than for the paretic side (11.7°).

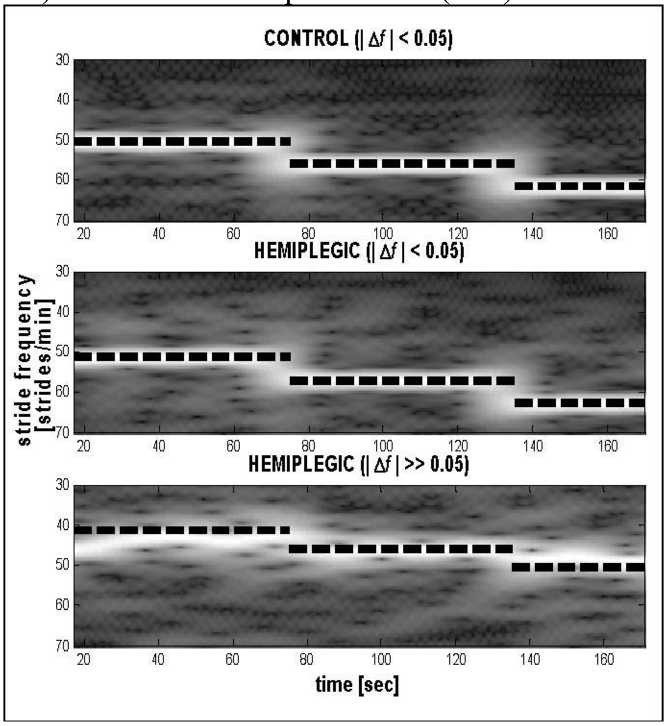

Figure 1. Spectrograms; contribution of frequency components is proportional to the brightness.

Discussion

The analysis of auditory-motor coordination showed that in stroke patients the relative phase variability between ipsilateral pacing signals and heel strikes σ was significantly lower for the non-paretic side than for the paretic side. In general, local decrease of variability may indicate points or regions in the workspace where useful task-specific information is available, such as information about the required timing [3]. Given such 'anchoring' [3], it may be postulated that stroke patients coordinate their gait to the perceptual stimulus by predominantly controlling (i.e., synchronizing, timing) the movements of the non-paretic body side to the ipsilateral pacing stimulus. Apparently, this is the most efficient way to deal with their gait asymmetry in this perceptual-motor task (i.e., paced walking). In the control group, no indications for lateralized auditory-motor anchoring were observed. These results substantiate the suggestion by Wagenaar and Beek [2] that tweaking the perception-action coupling by means of external rhythms can improve the spatiotemporal organization of pathological gait. Interestingly, from these observations new rehabilitation strategies might be distilled. For example, instructing hemiplegics to couple the heel strikes of their paretic instead of their non-paretic leg to the pacing signal might tentatively lead to improved gait coordination by increasing step length of the paretic limb.

References

[1] Wagenaar RC, Beek WJ (1992). Hemiplegic gait: a kinematic analysis using walking speed as a basis. Journal of Biomechanics, 25, 1007-1015

[2] Thaut MH, McIntosh GC, Rice RR (1997). Rhythmic facilitation of gait training in hemiparetic stroke rehabilitation. Journal of the Neurological Sciences, 151:207-212.

[3] Beek PJ (1989). Juggling dynamics. Free University Press, Amsterdam