order to evaluate the data for both clinical and research purposes, the potential magnitude of error in joint angle measurements needs to be known. Stereo-photogrammetric movement analysis is affected by three main sources of error: instrument errors, skin movement artifacts and anatomical landmark uncertainty. This study evaluates this latter source assessing the reliability of both intra and inter-examiner anatomical landmarks identification.

Methodology

Two healthy subjects were investigated. Each subject wore four skin marker clusters: on the pelvis, on the left thigh, shank and foot. Each cluster was made of three not rigidly connected markers conveniently located in front of two cameras of a stereo-photogrammetric system (ELITE, B.T.S. Milan). In order to assess intra-examiner reliability, the examiner (the Gait Laboratory physical therapist) was asked to identify a sequence of palpable anatomical landmarks following the directions outlined by Cappozzo et al. (1994). This sequence included Left and Right, Anterior and Posterior, Superior Iliac Spines (LASIS, RASIS, LPSIS, LPSIS), Greater Trochanter (GT), Medial and Lateral femural Epicondyles (ME, LE), Tibial Tuberosity (TT), Head of the Fibula (HF), Medial and Lateral Malleoli (MM, LM), CAlcaneus posterior surface (CA), dorsal aspects of First, Second and fifth Metatarsal head (FM, SM, VM). Additionally, the position of the Femur Head (FH) was assessed referring to the acetabulum center, assessed according to Bell, et al. (1990), in the femur reference frame during standing. According to the CAST protocol proposed by Cappozzo et al. (1995), a stick supporting two markers was used to point at each anatomical landmark, and a short static acquisition was performed. This anatomical landmark pointing procedure (calibration) was done six times consecutively. At no time did the subject change his/her marker set up. In order to assess inter-examiner reliability, six registered physical therapists working at Spaulding Rehabilitation Hospital in gait rehabilitation, were asked to perform once the same anatomical landmark calibration on the two subjects.

A mean anatomical reference was determined for each bone by averaging the anatomical landmark technical positions among the six calibrations: the Y axis in the cranio-caudal direction, the Z axis medio-lateral and the X axis frontward.

For each bone, the standard deviation of the orientation vectors (Spoor and Veldpaus, 1980) of the actual anatomical frame with respect to the mean anatomical frame, among the six calibrations, was assessed. Additionally, the joint orientation error was estimated by assessing the standard deviation, among the six calibrations, of the joint angles.

Results

In table 1 the anatomical landmark local positions standard deviation referred to the relevant bone mean anatomical reference frame of both the subjects and for both intra and inter-examiner tests are reported. The standard deviation of the bone orientation errors and the joint orientation errors are reported in table 2.

Table 1

		intra #1			intra #2			inter #1			inter #2		
	mm	X	Y	Z	X	Y	Z	X	Y	Z	X	Y	Z
PELVIS	LASIS	2.8	4.1	18.2	9	3.6	22.3	7.7	11	16.7	6.5	3.3	22.1
	RASIS	8.5	15.6	23.5	8.7	17	19.2	19.9	16.3	25.6	17.4	33.2	25
	LPSIS	6.6	6.5	20.7	8.7	4.6	19.6	17.5	12.5	21	7.9	9.9	22.7
	RPSIS	8.4	11.6	23.6	8.5	12.5	36.2	16.8	12.5	11.1	10.5	20.1	20.3
THIGH	LGT	9.2	9.1	7.5	12.4	9.8	17.6	16.4	17.2	12.9	18.4	26	19.9
	LME	6	13.6	20.1	16.7	9.2	20.2	8.2	13	10.4	10.7	8.2	21.2
	LLE	6.3	8.2	11.4	5.9	12.4	13	8.3	14	13.4	28.4	20	21
	LFH	7.2	9.5	13.9	20.7	12	17.3	16.8	11.7	10.2	13.8	12.3	13.5
TIBIA	LTT	13.8	4.5	13.2	2.9	1.9	9.3	3.1	7	26.2	3.4	9.3	11.7
	LHF	9.4	8.4	10.4	15.1	3.2	6.1	4.3	7.8	7.3	7.4	11.6	15
	LMM	10.8	6	12.5	14.4	15.1	13.5	11.1	8.7	16.6	11.5	9.5	26.3
	LLM	4.5	3.5	5	8.3	4.1	7.8	19.9	4.8	15.4	12.4	12.1	18.3
FOOT	LCA	12.8	14.8	29.2	13.1	7.4	16.8	10.8	15.7	19.5	9.5	8.1	14.2
	LFM	8.8	6.4	17.4	12.4	8.2	15.1	11.7	3.1	26.9	21.5	14.9	30.2
	LSM	5.5	6.6	25.4	7.1	6	23	9.6	8.1	18.1	14.5	4.9	30.4
	LVM	3.1	5.6	10.3	3.1	3.7	15.2	5.8	9.3	16.6	9.4	7.4	18.8

Table 2

	intra #1				intra #2				inter #1				inter #2			
deg	θ_x	θ_y	θ_z	abs(θ)	θ_x	θ_y	θ_z	abs(θ)	θ_x	θ_y	θ_z	abs(θ)	θ_x	θ_y	θ_z	abs(θ)
PELVIS	3	1.4	3.9	5.1	3.7	1.8	4.8	6.3	3.8	3.8	3.6	6.4	6.6	3.6	4.6	8.8
THIGH	1.9	1.5	0.8	2.5	4.3	9.9	3.5	11.4	1.9	4.8	1.7	5.4	3	12.8	4.3	13.9
SHANK	3.4	5	1.2	6.1	3.4	15.2	4.1	16.1	4.5	13.9	3.6	15	3.9	6.9	1.7	8.1
FOOT	14.4	7	4.4	16.6	12.2	17.3	7.8	22.5	8.2	11.4	4.1	14.6	3.5	18.5	6.2	19.8
HIP	3.8	2.7	4.3	6.4	6.2	10.8	7.7	14.7	2.8	6.3	4	8	8.8	10.3	6.6	15
KNEE	4.2	5.6	1.3	7.1	6.4	16.5	7.1	19	5.5	16.4	4.6	17.9	5.4	15.6	5.9	17.5
ANKLE	15.4	7.7	4	17.7	9	22.1	10.8	26.2	10.7	18.9	5.2	22.3	5.8	24.8	5	26

Discussion

The anatomical landmark identification error is greater than the other sources of errors. The intra-examiner tests showed errors smaller than those obtained in the inter-examiner tests. Among the body segment anatomical reference frames, the one of the foot is the most difficult to locate. The joint angles errors often have a standard deviation greater than ten degrees This is likely to cause very low reliability in assessing joint rotation offsets Additionally, these errors introduce a "kinematic crosstalk" among the angular components of joint kinematics that can significantly affect the smallest angular components (specially internal/external rotations), mainly in the case of mostly cylindrical joints.

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Validation of a method for location of the Hip Joint Center

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Introduction

The commercially available software package Orthotrak (Motion Analysis Corp., Santa Rosa, CA) uses a method of locating the hip joint center (HJC) which is based upon displacements along the x (anterior-posterior), y (medial-lateral), and z (superior-inferior) axes of the pelvic coordinate system. The magnitude of these displacements are determined from fixed percentages of the distance between the anterior superior iliac spines (ASIS) of the pelvis. Controversy has existed regarding the percentages which are appropriate for use in this method. Documentation for the original version of Orthotrak suggested values of -4% in the x direction, -32% in the y direction, and -34% in the z direction. These suggested values were based upon measurement of 31 skeletons. The current version of Orthotrak suggests values of -21%, -32%, and -34% for x, y, and z, respectively, which are considerably different. The stated reasons for the change in the suggested percentages are to account for marker diameter and soft tissue.

Given the large difference in values between the old and new x percentages recommended by the vendor, and the controversy surrounding the proper values, a study was conducted to determine which recommendations were more appropriate for use. The study was based on the model of the hip joint as a perfect ball and socket joint. We modeled a point rigidly fixed on the femur as a point moving at a constant radius, i.e., a point moving on a sphere, and solved for the center of this sphere thus finding the location of the hip center.

Methodology

Three-dimensional motion captures were taken of four normal subjects circumducting both right and left hips. Five trials were taken of the subject circumducting each hip. The subjects were outfitted with external markers on both ASIS, sacrum, a leg array (three markers rigidly attached to each other taped to the thigh), and a marker on either side of the knee axis. The data was tracked, and then analyzed in Orthotrak. Using Orthotrak, a virtual knee center for each data point was calculated. The virtual knee centers were the data points which were used to define the surface of the sphere which the hip joint is centered upon. The hip center was determined through minimization of the following error equation:

$$e = \frac{1}{n}\sum_{i=0}^{i=n}\left((x_i - x_0)^2 + (y_i - y_0)^2 + (z_i - z_0)^2 - R^2\right)^2$$

where e is the error (the quantity to be minimized) n is the total number of data points collected for the trial, x_0, y_0, z_0 are the x, y, and z coordinates of the center of the sphere (HJC location), x_i, y_i, z_i are the x, y, and z coordinates of the *ith* data point (location of the virtual knee center), and R is the radius of the sphere (femur length). The location of the HJC was determined for each trial and expressed as a percentage of the inter-ASIS distance.

The minimization problem was solved using a downhill simplex method available in the MATLAB software package (MathWorks, Inc., Natick MA). The values of x_0, y_0, z_0 and R which minimized the error are reported by this method.

Results

The results are shown in the following table. The values are the average of the five circumduction trials (except for subject 4, right hip, where only four trials were averaged).

Subject	X percentage	Y percentage	Z percentage
1R	-23.2	-35.0	-33.0
1L	-23.3	-34.9	-33.3
2R	-26.6	-36.1	-32.8
2L	-22.8	-32.8	-35.9
3R	-24.5	-43.4	-52.4
3L	-24.3	-37.3	-48.5
4R	-19.3	-43.2	-64.4
4L	-19.2	-41.7	-59.6
Mean	-22.8	-38.0	-45.0
Standard Deviation	2.5	4.13	12.9

Discussion

The objective of this study was to determine which of the percentages, the newer or older, was more appropriate. The percentage of greatest interest was the x coordinate, as this parameter had been changed from the original recommendation. In addition, there are disagreements among other motion labs as to which x percentage should be used. The previous percentage had been -4%, while the new percentage was -22%. This study indicates that the actual location of the hip center is much closer to the new percentage. The x percentage was the most consistent of the three measurements, with a mean of -22.8%, and a standard deviation of 2.5%. This supports -22% as a better estimate than the original recommendation.

The study indicates that the y and z percentages are considerably more variable across subjects than the x percentage. The Orthotrak percentages for y and z are -32% and -34% respectively. The values from the first two subjects fall near these values. In the last two subjects, the y percentage is higher, and the z percentage is almost twice that of the suggested value. The y and z values obtained in this preliminary study are inconsistent with either of the Orthotrak recommendations. A larger sample is necessary to further validate the percentages obtained. However, the results obtained, to date, support the percentages suggested most recently by Motion Analysis Corp. as an improvement over the older recommendations.

Error in Kinematic Data Secondary to Excesses in Femoral Torsion

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Introduction

Bony torsion may alter kinematic measurements leading to clinical misinterpretation. Kinematic models used in gait analysis may be based on faulty assumptions of normal femoral or tibial torsion. Some of the patient groups that use gait analysis the most, e.g., patients with cerebral palsy, often have one or more rotational deformities. It is important for any clinician reviewing kinematic data to realize how the kinematic model being used handles these excessive bony torsions.

Some models reportedly correct for the effects of bony torsions on joint rotation angles. Mann et al., 1983 (the MIT model as described by Giannini et al., 1994), describes taking static pictures and subtracting these static calculations from the dynamic calculations to obtain the dynamic or corrected joint rotation angles. Kadaba et al., 1990 (Helen Hayes model), also seems to correct by aligning a lateral thigh wand with the knee axis and long axis of the femur to define neutral rotation angles while standing in a normal position. These correction procedures should result in situations where, for example, the hip internal/external rotation angles of a patient with femoral anteversion shows little or no internal rotation when, visibly, the knee is internally rotated by clinical standards. Davis et al., 1991 (Newington model), and the Motion Analysis Corporation, 1993 (Cleveland Clinic model), report correcting for excessive tibial torsion by shifting marker placement of the lateral malleolus, or both malleoli, based on torsion characteristics of the tibia; this correction will have a similar effect. Tylkowski et al. (1982) corrects for femoral and tibial torsion by aligning one of the axes of the reference frame in each segment parallel to the Y-axis of the global reference frame. The axis formed by an anterior wand on the thigh is aligned with the posteroanterior axis of the global reference frame while the subject is standing. Thus, if the knee axis is internally rotated due to femoral anteversion, the kinematic measurements will again report only actual motion rather than the rotated alignment.

Ultimately, important clinical decisions, including surgical interventions, orthosis or therapy prescriptions, based on results of kinematic data must be made with a clear understanding of the effects of excessive bony torsions on the kinematic data. If the data were adjusted or corrected, the implications of this also must be understood. Furthermore, the effects of correcting the kinematic data or not correcting the data on kinetic parameters should be investigated.

The purpose of this work was to investigate the effects of femoral torsion on hip kinematics when using any model which uses the knee axis to construct the thigh reference frame. Kinematic data from these types of models would be affected by excesses of femoral torsion since femoral torsion manifests itself at the distal end of the femur (Gage, 1991).

Methodology

Kinematic data from the right and left hips of one adult woman without disabilities was used. The data were altered to simulate effects of femoral anteversion and retroversion by rotating the thigh's reference frame, in increments of 5 degrees, around the long axis of the thigh (from knee to hip center). Rotating the reference frame has the same effect as an externally or internally rotated knee axis secondary to excessive femoral torsion. The reference frame was rotated internally a total of 45 degrees to simulate a more anteverted femur, and externally a total of 15 degrees to simulate a more retroverted femur. Less retroversion was simulated since populations with rotational deformities are femorally anteverted more often than retroverted.

Results

Rather than a clear offset of the neutral, zero point, of the hip rotation angles calculated, there was a progressive increase in both internal and external rotation as anteversion is increased, and a progressive decrease in both with retroversion. The total average error of the data from the right and left hips was 1.5 degrees throughout the range of motion with each 5 degree increase in femoral torsion. This represents a 10% error based on norms of maximum hip internal/external range of motion reported by Perry (1992) during walking (15 degrees).

Discussion

Any model which uses points on the knee axis (Mann et al., 1983; Motion Analysis Corporation, 1993; Giannini et al., 1994), wands (Apkarian et al., 1989) or clusters (Davis et al., 1991) aligned with the knee axis to construct the frontal plane of the femur will be affected in the way that we have simulated here. Since the effect of excessive torsion on the model resulted in an unusual offset, people using other types of kinematic models should investigate the errors that excessive femoral torsions will cause using their model. Ultimately, a decision has to be reached as to how to deal with the kinematic error found. This raises important issues. If the kinematic error is corrected, as is done by Giannini et al. (1994), Kadaba et al. (1990) and Tylkowski et al. (1982), and these corrected data are used in the calculations of kinetic parameters, it seems that error will be introduced into these data. However, if the kinematic data are reported without correction, it is likely that these data will be misinterpreted as motions rather than torsions. Hence, our recommendation is that the data be collected without correction, and that these data be used for the calculation of kinetic parameters. Kinematic data should either be corrected before they are graphed for clinical interpretation or presented without correction. However, an assessment of torsion must be included in the gait report, and the method of display (corrected or uncorrected) must be explicitly stated.

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Session 13: Functional Assessment and Gait

The Relationship of Muscle Strength and Gait Performance in Children with Juvenile Idiopathic Inflammatory Myopathy

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Introduction

In 1993, the National Center for Medical Rehabilitation Research within the National Institutes of Health identified a need for research which relates physical impairments to the functional limitations that they produce. Toward this goal, several researchers have investigated the relationship between muscle strength and walking speed in adults with a variety of medical diagnoses in an effort to identify which muscles predict gait performance. The purpose of the present study was to reexamine the relationship of lower extremity muscle strength with walking speed, stride length, and gait cycle time in a group of children with juvenile idiopathic inflammatory myopathy (JIIM) using two analytic approaches to characterize muscle strength.

Methodology

Twenty-two children (16 girls and 6 boys) ranging in age from 5 to 18 years old (mean=11.4) and diagnosed with JIIM participated in the study. Manual muscle testing (MMT) using the Kendall 10 point scale was performed bilaterally on the hip flexors, abductors, and extensors, the knee extensors, and the ankle dorsiflexors and plantar flexors. Gait analysis was performed with a 6-camera Vicon system and two force plates while children walked barefoot at a self-selected speed. Measured gait parameters included stride length, gait cycle time, and walking speed. Stride length and walking speed were scaled by subject height to reduce the effect of age on these measures.

Periodic repeat evaluations were performed and included in the statistical analysis only if results differed from the previous test. The final analysis included 33 evaluations from 22 subjects. A multiple regression analysis with forward stepping was performed to predict either walking speed, stride length, or gait cycle time. In the first approach, the independent variables in the regression analysis were the raw MMT scores of each tested muscle (score of the stronger hip flexor, weaker hip flexor, stronger hip abductor ...through... score of the weaker ankle plantar flexor). In the second approach, the independent variables in the regression analysis were the total number of muscles that tested weaker than each MMT strength grade (number of muscles weaker than 10/10, number weaker than 9/10 ...through... number weaker than 1/10).

Results

MMT scores ranged from 1/10 to 10/10, were lower in the proximal muscles than in the distal muscles, and were asymmetric by one half grade or less. On average, subjects walked with a stride length of 0.76 statures (range 0.33 to 0.97) and a gait cycle time of 1.02 seconds (range 0.74 to 1.64), resulting in a walking speed of 0.75 statures/s (range 0.20 to 1.17). The results of the regression analyses (Table) revealed that approximately 65% of the variability in walking speed, stride length, and gait cycle time could be predicted from either method used to characterize lower extremity strength (raw MMT scores of each muscle or number of muscles weaker than each strength grade).

Table: Results of Forward Regression Analysis (p < 0.01 for all analyses)

	Raw MMT Scores of each Muscle		*Number of Muscles Weaker than each MMT Strength Grade*	
	Final Model included:	Results	Final Model included:	Results
Walking Speed	Stronger Hip Flexors and Stronger Ankle Dorsiflexors	R =0.83 R^2=0.69	Number of Muscles less than 6/10	R =0.80 R^2=0.64
Stride Length	Stronger Hip Flexors	R =0.80 R^2=0.63	Number of Muscles less than 6/10 and less than 7/10	R =0.82 R^2=0.67
Gait Cycle Time	Stronger Hip Flexors and Stronger Ankle Dorsiflexors	R =0.79 R^2=0.62	Number of Muscles less than 6/10	R =0.82 R^2=0.67

Discussion

Results of this study revealed that gait performance as represented by walking speed, stride length, and gait cycle time could be predicted from lower extremity muscle strength in children