ELSEVIER

Journal of Biomechanics 34 (2001) 437–447

JOURNAL OF BIOMECHANICS

www.elsevier.com/locate/jbiomech
www.JBiomech.com

Rotational moment arms of the medial hamstrings and adductors vary with femoral geometry and limb position: implications for the treatment of internally rotated gait

Allison S. Arnold*, Scott L. Delp

Biomechanical Engineering Division, Mechanical Engineering Department, Stanford University, Stanford, CA 94305 3030, USA

Accepted 6 December 2000

Abstract

Persons with cerebral palsy frequently walk with a crouched, internally rotated gait. Spastic medial hamstrings or adductors are presumed to contribute to excessive hip internal rotation in some patients; however, the capacity of these muscles to produce internal rotation has not been adequately investigated. The purpose of this study was to determine the hip rotation moment arms of the medial hamstrings and adductors in persons with femoral anteversion deformities who walk with a crouched, internally rotated gait. A musculoskeletal model with a "deformable" femur was developed. This model was used, in conjunction with kinematic data obtained from gait analysis, to calculate the muscle moment arms for combinations of joint angles and anteversion deformities exhibited by 21 subjects with cerebral palsy and excessive hip internal rotation. We found that the semimembranosus, semitendinosus, and gracilis muscles in our model had negligible or *external* rotation moment arms when the hip was internally rotated or the knee was flexed — the body positions assumed by the subjects during walking. When the femur was excessively anteverted, the rotational moment arms of the adductor brevis, adductor longus, pectineus, and proximal compartments of the adductor magnus in our model shifted toward external rotation. These results suggest that neither the medial hamstrings nor the adductors are likely to contribute substantially to excessive internal rotation of the hip and that other causes of internal rotation should be considered when planning treatments for these patients.

Keywords: Musculoskeletal model; Hip; Anteversion; Gait; Cerebral palsy

1. Introduction

Children with cerebral palsy frequently walk with excessive internal rotation of the hip. Spastic medial hamstrings or adductors, among other factors, are thought to contribute to the excessive internal rotation in many patients based on electromyographic evidence that the muscles are active during walking, and the presumption that these muscles generate an internal hip rotation moment (e.g., Sutherland et al., 1969; Chong et al., 1978). Surgical lengthening of the semimembranosus, semitendinosus, adductor brevis, adductor longus, and/or gracilis is often expected to decrease excessive internal rotation (Hoffer, 1986; Root, 1987; Tachdjian, 1990). However, the extent to which the hamstrings and adductors contribute to hip internal rotation is unclear, and the changes in hip rotation following surgery are inconsistent.

The rotational moment arm of a muscle about the hip determines whether the muscle has the potential to produce an internal or an external hip rotation moment. Previous studies of hip rotation moment arms — based on adult-sized cadavers and computer models that represent normal musculoskeletal geometry — have indicated that the medial hamstrings, adductor brevis, and adductor longus have small (1 cm or less) internal rotation moment arms in the upright, standing position (Dostal et al., 1986; Mansour and Pereira, 1987; Lengsfeld et al., 1997). These data have provided insight into the rotational capacity of the muscles for a limited set of body positions in unimpaired adults. It remains unclear, however, whether such data accurately characterize the actions of muscles in children with neuromuscular disorders.

*Corresponding author. Tel.: +1-650-724-7086; fax: +1-650-725-1587.

E-mail address: asarnold@stanford.edu (A.S. Arnold).

0021-9290/01/$ - see front matter
PII: S0021-9290(00)00232-3

Descriptions of hip rotation moment arms for unimpaired subjects in the upright position may be misleading when analyzing rotational abnormalities in persons with cerebral palsy for two main reasons. First, rotational abnormalities of the hip are often accompanied by excessive anteversion of the femur (Bleck, 1987), a torsional bone deformity that may alter the lines of action and moment arms of muscles about the hip. Second, the moment arms must be evaluated over the range of limb positions assumed by persons with cerebral palsy during walking; this frequently includes exaggerated flexion of the hips and knees in addition to increased internal rotation of the hip. Without such moment arm data, the potential of the hamstrings or adductors to produce an internal rotation moment in these patients is unknown, and a scientific rationale for lengthening these muscles to treat excessive internal rotation of the hip does not exist. We believe that a detailed analysis of the muscle moment arms, which considers variations in femoral geometry and limb position, is needed to improve the planning of muscle–tendon surgeries intended to reduce internal rotation.

As a first step toward this goal, we evaluated the hip rotation moment arms of the medial hamstrings and adductors using highly accurate musculoskeletal models of three individuals with cerebral palsy that we constructed from magnetic resonance (MR) images. Analysis of these models, at the limb positions corresponding to each subject's internally rotated gait, revealed that the semimembranosus, semitendinosus, adductor brevis, adductor longus, and gracilis had external rotation moment arms or very small internal rotation moment arms throughout the gait cycle, in all three subjects (Arnold et al., 2000a).

Based on these observations, we hypothesized that the rotational moment arms of the medial hamstrings and adductors are shifted toward *external* rotation by excessive femoral anteversion and/or by exaggerated hip flexion, knee flexion, or hip internal rotation. We tested this hypothesis using a musculoskeletal model with a "deformable" femur that estimates the muscle moment arms at the body positions of patients who walk with crouched, internally rotated gait.

2. Materials and methods

A computer model of the lower extremity with a "deformable" femur was developed. This model characterizes the geometry of the pelvis, femur, and proximal tibia, the kinematics of the hip and tibiofemoral joints, and the paths of the medial hamstrings and adductor muscles for an average-sized adult male. This model is similar to the graphics-based, deformable lower limb models, we have used in previous studies (Arnold et al., 1997; Schmidt et al., 1999), with the following notable improvements. These improvements were motivated, in large part, by our desire to accurately estimate the rotational moment arms of the muscles for a wide range of joint angles and anteversion deformities. First, we refined the locations of the muscle attachments reported by Delp et al. (1990) to more closely correspond to three-dimensional surface representations of the muscles and bones generated from approximately 250 MR images of each of three lower extremity cadaveric specimens. Second, we implemented a description of tibiofemoral kinematics that accounts for the three-dimensional rotations and translations of the tibia relative to the femur (Walker et al., 1988); in previous models, we neglected the rotations of the tibia in the frontal and transverse planes. Third, we defined "wrapping surfaces" (van der Helm et al., 1992), in addition to "via points" (Delp et al., 1990), to characterize the curved paths of the muscles with hip internal rotation and other joint motions. The wrapping surfaces were designed to simulate interactions between the muscles and underlying structures, thereby providing an improvement over previous models that estimated rotational moment arms based on straight-line approximations of the muscle-tendon paths (Dostal et al., 1986; Lengsfeld et al., 1997; Delp et al., 1999). Lastly, we developed new algorithms to alter the geometry of the proximal femur. These algorithms were based on careful inspection of the deformed femurs of four individuals with cerebral palsy constructed from MR images. Our resulting model was capable of estimating the hip rotation moment arms of the medial hamstrings and adductors for a 60° range of femoral anteversion angles and a variety of body positions, including hip and knee angles that corresponded to crouched, internally rotated gait. The following paragraphs further describe our efforts to develop, test, and analyze this model.

We defined the bone geometry, joint kinematics, and muscle-tendon paths of our deformable model using a musculoskeletal modeling package, SIMM (Delp and Loan, 1995). The surface geometry of each bone was described by a polygonal mesh. Coordinate systems for the pelvis, femur, and tibia were established from anatomical landmarks (Arnold et al., 2000b), and kinematic descriptions of the hip and tibiofemoral joints were specified based on the bone surface geometry. The hip was represented as a ball-and-socket joint. The tibiofemoral joint prescribed the translations and rotations of the tibia relative to the femur as functions of knee flexion angle, and was based on published experimental measurements of tibiofemoral kinematics (Kurosawa et al., 1985; Nisell et al., 1986; Walker et al., 1988). Our procedures for establishing the segment coordinate systems and joint kinematics have been reported in detail previously (Arnold et al., 2000b).

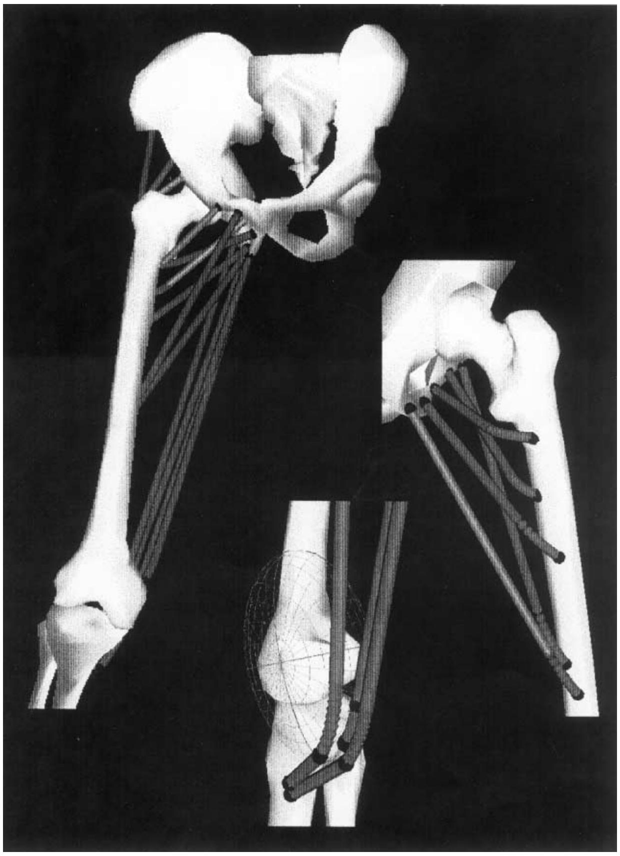

Fig. 1. Representation of muscle–tendon paths. Ellipsoidal wrapping surfaces were positioned at the distal femur to prevent the medial hamstrings and gracilis from penetrating the posterior femoral condyles and/or adjacent soft tissues with knee extension (*center*). Via points were added proximal to the insertions of the semitendinosus and the gracilis to simulate the constraints produced by surrounding tissues. Wrapping surfaces were distributed along the shaft of the femur to characterize the wrapping and sliding of the adductor brevis, adductor longus, adductor magnus, and pectineus over the femoral shaft and underlying muscles with hip flexion and internal rotation (*right*).

The paths of the semimembranosus, semitendinosus, gracilis, adductor brevis, adductor longus, adductor magnus (proximal, middle, distal, and ischiocondylar compartments), and pectineus muscles were defined for a range of hip and knee motions. Each muscle was represented as a series of line segments from origin to insertion. The attachment sites of the muscles were identified, and wrapping surfaces and via points were added to simulate underlying structures and other anatomical constraints (Fig. 1). We determined the muscle attachment sites by graphically superimposing three-dimensional surface meshes of the muscles and bones, generated from MR images of three lower extremity cadaveric specimens, onto our deformable model. We prescribed the paths of the muscles through a range of hip and knee motions by specifying wrapping surfaces (for all muscles) and via points (for semitendinosus and gracilis) as follows. First, for each of the three specimens, we created a graphics-based kinematic model of the hip joint, the tibiofemoral joint, and the surrounding musculature from MR images (Arnold et al., 2000a, b). Second, for each muscle–tendon compartment, we developed an algorithm to specify the position, orientation, and dimensions of an ellipsoidal wrapping surface and the locations of via points relative to skeletal landmarks. We chose landmarks that could be identified on each of the MR-based models and on the deformable model. We defined each algorithm such that the muscle moment arms calculated with the MR-based models compared favorably to the moment arms determined experimentally on the same specimens from tendon excursion experiments (Arnold et al., 2000a, b). We calculated the moment arms from the models using the partial velocity method (Delp and Loan, 1995). Once an algorithm was developed that could predict the moment arms with sufficient accuracy for all three specimens, the same algorithm was used to specify the muscle-tendon paths of our deformable model.

We developed techniques to deform the femur of our model to represent excessive anteversion and other deformities commonly observed in persons with cerebral palsy who walk with an internally rotated gait. Anteversion of the femur is defined as the angle between the plane of the femoral neck axis and the plane of the condylar axis (Fig. 2). This angle is generally between 10 and 20° in adults with normal femoral geometry (Upadhyay et al., 1987; Miller et al., 1993), but may be 60° or more in children with cerebral palsy (Ruwe et al., 1992; Laplaza et al., 1993). The femur of our undeformed model has an anteversion angle of 20°.

In previous efforts to develop a musculoskeletal model with a "deformable" femur, we and other investigators have typically assumed that anteversion deformities occur entirely within the femoral neck, superior to the lesser trochanter (Arnold et al., 1997; Schutte et al., 1997; Schmidt et al., 1999). We tested this assumption in this study. Examination of the deformed femurs of four subjects with cerebral palsy, generated from MR images, revealed that previous assumptions about anteversion deformities were not entirely accurate. The femoral neck was deformed in these subjects, but the lesser trochanter and proximal femoral shaft also appeared to be involved. This finding is consistent with a recent report by Lundy et al. (1998), who inspected the femurs of 12 patients undergoing proximal femoral resection.

Based on these observations, we altered the femoral anteversion angle of our deformable model by rotating and/or translating the bone vertices that make up the femoral head, neck, and shaft (Fig. 3; also visit the webpage of the Journal of Biomechanics: http://www.elsevier.nl:80/inca/publications/store/3/2/1/ to view animations of the deformable model "in action"). Two successive transformations were

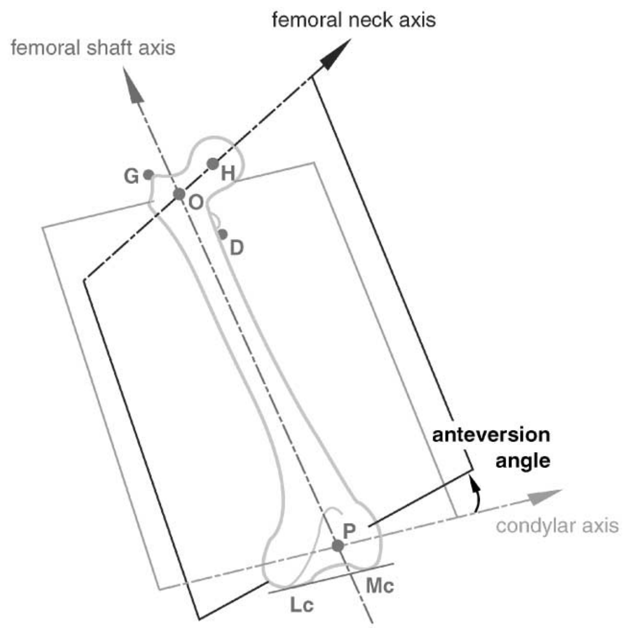

Fig. 2. Geometric description of femoral anteversion angle. H is the center of the femoral head. G is the most superior point on the greater trochanter. D is the most distal point on the lesser trochanter. P is the attachment of the posterior cruciate ligament. O is the center of the base of the femoral neck, which was determined by iteratively locating the centroid of the femoral diaphysis on a cross section through the midpoint of the vector joining points G and D, perpendicular to the vector joining points O and P. Lc and Mc are the posterior aspects of the lateral and medial condyles. The femoral neck axis is defined by points O and H, the femoral shaft axis by points O and P; these two axes define the plane of the femoral neck. Anteversion is the angle formed by the plane of the femoral neck and the plane of the condylar axis, which passes through points O and P parallel to the vector joining points Lc and Mc. Figure adapted from Murphy et al. (1987).

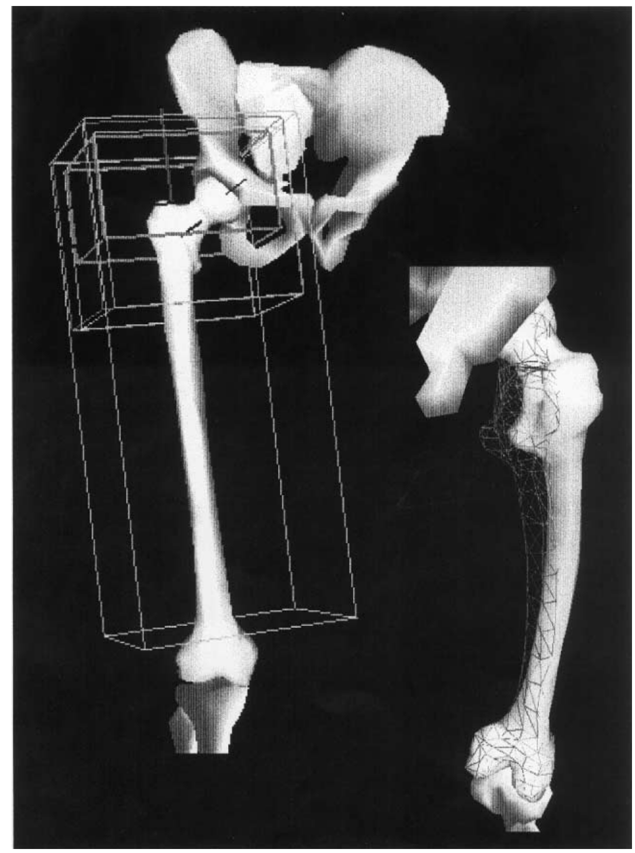

Fig. 3. Algorithm for deforming the femur. Anteversion deformities were represented in the model by rotating and/or translating the bone vertices that make up the femoral head, neck, and shaft. Vertices comprising the femoral head and neck (*within the inner box*) were rotated anteriorly about the femoral shaft axis through an angle corresponding to the desired change in anteversion. Vertices in the transition region, encompassing the lesser trochanter and proximal femoral shaft (*between the inner and middle boxes*), were rotated anteriorly about the femoral shaft axis through an angle that was decreased linearly as a function of superior–inferior distance along this axis. Vertices comprising the entire femur proximal to the femoral condyles (*within the outer box*) were translated after each deformation to restore the position of the femoral head to the acetabulum (*compare undeformed shaded bone and deformed wireframe bone*). The position of the knee center with respect to the hip center was not changed.

performed to represent an anteversion deformity. First, we rotated the femoral head, neck, and proximal shaft (i.e., vertices comprising the femur proximal to and including the lesser trochanter) anteriorly about the femoral shaft axis, thereby increasing the angle between the plane of the femoral neck axis and the plane of the condylar axis. Second, we translated the bone vertices proximal to the femoral condyles to restore the position of the femoral head in the acetabulum. Muscle insertions on the femur were displaced with the bone vertices. Hence, the lines of action and moment arms of the adductor brevis, adductor longus, adductor magnus, and pectineus muscles were altered by these deformities. The position of the knee center with respect to the hip center was not changed; thus, the paths of the medial hamstrings and gracilis were not affected. This model is the first to provide a mathematical description of excessive anteversion that captures the salient features of deformed femurs constructed from MR images.

We examined the accuracy with which our deformable model could estimate the rotational moment arms of the medial hamstrings and adductors as follows. First, we created graphics-based kinematic models of three subjects with cerebral palsy from MR images (Arnold et al., 2000a). Second, we adjusted the femoral anteversion angle of our deformable model to match the anteversion angle of each subject; i.e., we deformed the model such that the angle between the plane of the femoral neck axis and the plane of the condylar axis was the same as the angle measured from the model of each subject. Third, we compared the rotational moment arms of the muscles estimated with these "deformed" models to the moment arms calculated from the MR-based models at the limb positions corresponding to each subject's internally rotated gait. The subjects

Table 1
Errors[a,b,c] in peak internal rotation moment arms during gait estimated with the deformable model

	Subject A	Subject B	Subject C
Semimembranosus	−1	1	0
Semitendiniosus	−2	1	0
Gracilis	−2	3	1
Adductor brevis	1	−3	−3
Adductor longus	1	−1	−3
Pectineus	3	1	3
Adductor magnus (proximal)	−3	−1	−4
Adductor magnus (middle)	0	1	−2
Adductor magnus (distal)	1	0	−2
Adductor magnus (ischiocondylar)	1	1	1

[a] Error defined as peak moment arm estimated with the deformable model — peak moment arm calculated with the MR-based model; the anteversion angle of the deformable model was adjusted to match the anteversion angle of each subject.

[b] Positive error value indicates that the moment arm estimated with the deformable model was more internal than the moment arm calculated with the MR-based model.

[c] Units of mm.

ranged in age from 7 to 27 years and had anteversion angles of 44–47°. The fourth subject with cerebral palsy who was imaged walked with external rotation of his hips and was excluded from this analysis.

Differences in the rotational moment arms of the medial hamstrings and adductors determined from the MR-based models and estimated with the deformed models were small (Table 1); errors in the peak internal rotation moment arms during walking were at most 4 mm (proximal compartment of adductor magnus, Subject C) over the gait cycles of the three subjects. The peak moment arms determined from the MR-based and deformed models occurred at similar times in the gait cycle. Based on these data, we believe our deformable model is suitably accurate for estimating the hip rotation moment arms of the medial hamstrings and adductors for joint angles and femoral geometries typical of persons who walk with excessive internal rotation of the hip.

We used our deformable model to evaluate our hypothesis in two steps. First, we used the model to determine how independent variations in hip rotation (−40° external rotation to 40° internal rotation), hip flexion (full extension to 90° flexion), knee flexion (full extension to 90° flexion), and femoral anteversion (0–60°) affect the rotational moment arms of the medial hamstrings and adductors. Second, we obtained gait analysis measurements for 21 subjects (23 limbs) with excessive internal rotation of the hip, and we used our model in conjunction with the subjects' measured gait kinematics and estimated femoral anteversion angles to calculate the muscle moment arms at the limb positions corresponding to each subject's abnormal gait. Our goal was to determine whether the medial hamstrings or adductors could produce an internal hip rotation moment in these subjects, or whether these muscles are likely to have external rotation moment arms for the ranges of anteversion angles and body positions that we observed.

The subjects who were analyzed in this study exhibited a wide variety of gait abnormalities (Table 2). Eighteen of the subjects had a diagnosis of spastic diplegia; three subjects had a diagnosis of spastic hemiplegia. All of the subjects were age seven or older, had no previous orthopaedic surgery, and walked without orthoses or other assistance. Nineteen of the subjects walked with 15° or more of hip internal rotation in only one limb; two subjects walked with excessive hip internal rotation in both limbs (Table 2, limbs 8–9, 12–13).

Each subject underwent gait analysis at the Children's Memorial Medical Center in Chicago. A five-camera motion measurement system (VICON, Oxford Metrics, Oxford, UK) was used to compute the subject's three-dimensional gait kinematics as described by Kadaba et al. (1990), based on estimates of the joint center locations as suggested by Davis et al. (1991). One representative trial for each internally rotated limb was selected, and the measured joint angles were transformed to be consistent with the coordinate systems that describe our deformable model (Arnold et al., 2000b). The anteversion angle of each subject was estimated based on the subject's measured range of passive hip internal and external rotation (Tables 2 and 3). The hip rotation moment arms of the medial hamstrings and adductors were calculated, at every 2% of the gait cycle, using the appropriately deformed model. The maximum internal rotation moment arm (or minimum external rotation moment arm) of each muscle over the gait cycle was noted. Our protocols for human subjects were approved by Institutional Review Boards at the Children's Memorial Medical Center and at Northwestern University. All subjects and/or their parents provided informed written consent.

Table 2
Characteristics of the cerebral palsy subjects

Limb	Gender	Age (yrs)	Average hip internal rotation[a] during gait	Hip flexion[b] during stance phase (max/min)	Knee flexion[c] during stance phase (max/min)	Hip Rotation[d] Range of Motion (internal/external)	Assumed Anteversion Angle[e]
1	M	8	15	46/14	43/34	40/30	20
2	F	10	20	62/13	37/18	50/30	30
3	M	13	16	50/12	26/9	55/25	30
4	F	10	17	58/26	74/69	65/30	30
5	M	13	22	56/27	62/55	50/15	30
6	M	7	15	40/−1	40/20	50/15	30
7-A[f]	F	7	31	51/3	33/3	65/25	40
8	F	10	25	52/42	77/73	70/30	40
9	F	10	24	58/30	64/58	70/30	40
10	M	7	19	44/7	48/30	50/5	40
11	M	12	22	45/24	65/54	80/30	40
12-C[f]	M	27	32	61/28	83/73	80/30	40
13	M	27	24	60/31	73/56	70/20	40
14	F	9	24	31/1	26/24	75/15	50
15	F	17	20	47/7	29/7	70/10	50
16	M	7	24	57/1	44/21	75/5	50
17	M	13	15	31/0	33/3	90/20	50
18	F	20	44	41/6	47/35	75/0	50
19	F	12	33	52/23	56/46	90/10	60
20	F	18	35	41/20	37/32	85/5	60
21-B[f]	M	14	37	39/17	39/26	90/10	60
22	F	15	39	56/29	38/30	75/−10[g]	60
23	M	7	33	44/0	40/20	90/0	60

[a] Rotation of the thigh in the transverse plane relative to the pelvis as tracked from a mid-thigh marker, in units of degrees; hip internal rotation is represented as a positive angle and is 0° at the anatomical position.
[b] Angle formed between the long axis of the thigh and a line perpendicular to the plane formed by the left and right anterior superior iliac spines and posterior superior iliac spines, in units of degrees; hip flexion is represented as a positive angle and is approximately 12° at the anatomical position.
[c] Angle formed between the long axis of the thigh and the shank, in units of degrees; knee flexion is represented as a positive angle and is 0° at the anatomical position.
[d] Passive range of hip rotation measured in the prone position with the hip extended and the knee flexed 90°; hip internal/external rotation ranges determined by the angle of the shank relative to vertical (Bleck, 1987), in units of degrees.
[e] Estimated from Table 3; in units of degrees.
[f] Limb for which an MR-based model was created.

3. Results

At the upright standing position, with normal anteversion of the femur, the medial hamstrings, adductor brevis, adductor longus, pectineus, and ischiocondylar compartment of the adductor magnus in our deformable model have hip rotation moment arms that are slightly internal (Fig. 4, shaded regions). The gracilis and the proximal compartment of the adductor magnus have hip rotation moment arms that are slightly external. The middle and distal compartments of the adductor magnus in our model have rotational moment arms that are negligible. These observations are consistent with our tendon excursion experiments (Arnold, 1999) and with EMG studies and moment arms published in the literature (Basmajian and DeLuca, 1985; Dostal et al., 1986; Lengsfeld et al., 1997).

Table 3
Asymmetry index used to estimate femoral anteversion angle

Asymmetry[a] in hip rotation range of motion	Assumed anteversion angle
$<20°$	20°
$\geqslant 20°$ but $<40°$	30°
$\geqslant 40°$ but $<60°$	40°
$\geqslant 60°$ but $<80°$	50°
$\geqslant 80°$	60°

[a] Asymmetry defined as range of internal rotation — range of external rotation.

When the hip is internally rotated more than about 20° (Fig. 4C), or the knee is flexed more than about 30° (Fig. 5), the rotational moment arms of the semimembranosus and semitendinosus in our model switch from internal to external rotation. The rotational moment

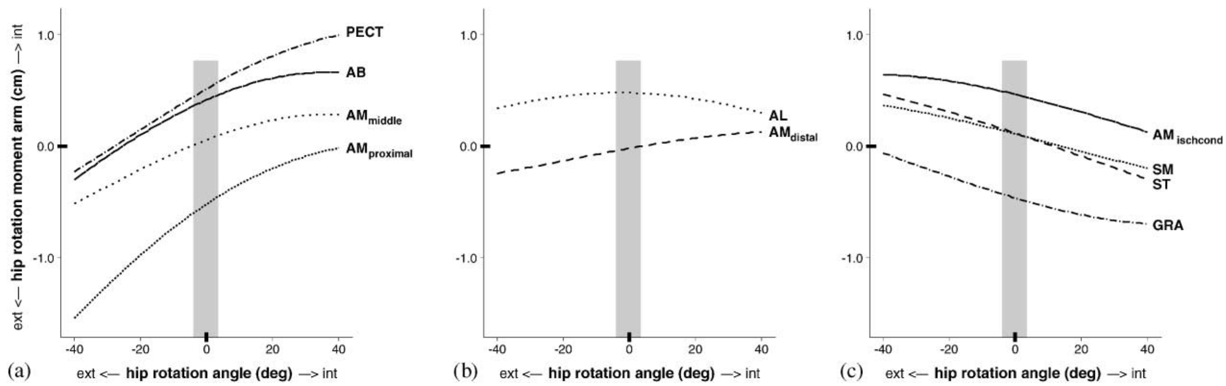

Fig. 4. Hip rotation moment arm vs. hip rotation angle for the semimembranosus (SM), semitendinosus (ST), gracilis (GRA), adductor brevis (AB), adductor longus (AL), adductor magnus (AM), and pectineus (PECT), calculated with the undeformed model. Hip flexion, adduction, and knee flexion angles are 0°. At the upright standing position (*shaded areas*), the pectineus, adductor brevis, adductor longus, medial hamstrings, and ischiocondylar compartment of the adductor magnus have hip rotation moment arms that are internal. The gracilis and the proximal compartment of the adductor magnus in our model have rotational moment arms that are external.

arm of the gracilis also becomes more external with hip internal rotation and knee flexion, and the moment arm of the ischiocondylar compartment of the adductor magnus becomes less internal with hip internal rotation.

Adductors that insert along the proximal femoral shaft have hip rotation moment arms that become more internal with hip internal rotation (Fig. 4A). With excessive femoral anteversion, however, the moment arms of the adductor brevis (Fig. 6), pectineus, and middle compartment of the adductor magnus in our model switch from internal to external rotation. Similarly, the moment arm of the adductor longus becomes less internal, and the moment arm of the proximal compartment of the adductor magnus becomes more external, with increased femoral anteversion.

Estimates of the muscle moment arms during crouched, internally rotated gait (Fig. 7) reflect these trends toward external rotation. For example, our analysis of the moment arms for 21 subjects with cerebral palsy (23 limbs) revealed that the semimembranosus, semitendinosus, and gracilis had hip rotation moment arms that were less than 1 mm or external throughout the gait cycle. This occurred because all of the subjects walked with excessive hip internal rotation, and many of the subjects also exhibited exaggerated knee flexion. The adductor brevis and the proximal, middle, and distal compartments of the adductor magnus had hip rotation moment arms, in most subjects, that were external throughout the gait cycle. Only one subject (Limb 1, Table 2) had an internal rotation moment arm of the adductor brevis during walking that was greater than 5 mm; this subject had an anteversion angle that was within the normal range. None of the subjects had an internal rotation moment arm of the adductor longus that was greater than 5 mm. The pectineus and the ischiocondylar compartment of the adductor magnus, in most subjects, had rotational

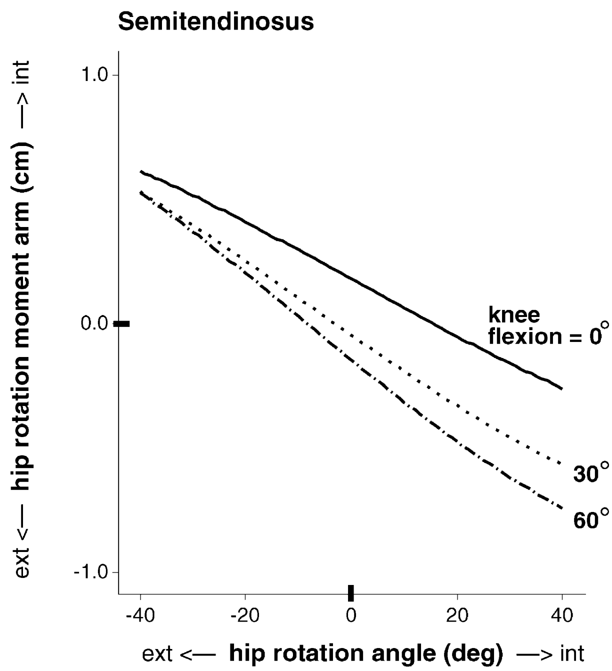

Fig. 5. Hip rotation moment arm vs. hip rotation angle for the semitendinosus, calculated using the undeformed model with 0, 30 and 60° knee flexion. Hip flexion and adduction angles are 0°.

moment arms that were internal throughout the gait cycle. However, neither of these muscles had an internal rotation moment arm that was greater than 8 mm during walking, for any of the subjects, based on our deformable model.

4. Discussion

The success of muscle–tendon surgeries to correct excessive internal rotation of the hip is limited, in part, because the rotational functions of the hip muscles — in children with femoral deformities, at the limb positions corresponding to walking — are not well understood. Only a few investigators have reported the hip rotation moment arms of muscles (Dostal et al., 1986; Mansour and Pereira, 1987; Lengsfeld et al., 1997; Delp et al., 1999; Arnold et al., 2000a); none have given a full account of how the muscle moment arms vary with hip rotation, hip flexion, knee flexion, and femoral anteversion. As a result, interventions commonly performed to treat crouched, internally rotated gait, such as lengthening of the medial hamstrings, adductor brevis, adductor longus and gracilis, are currently planned without quantitative descriptions of the muscle actions under appropriate biomechanical conditions. Identification of the muscles that have the potential to produce internal rotation moments about the hip is necessary to establish a rational basis for planning surgery.

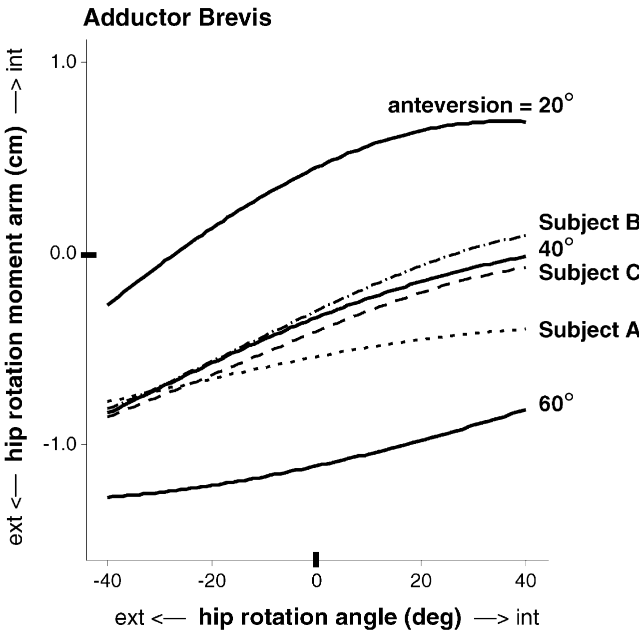

Fig. 6. Hip rotation moment arm vs. hip rotation angle for the adductor brevis, calculated using the deformable model with 20, 40 and 60° of anteversion (*solid lines*), and determined from MR-based models of three cerebral palsy subjects (*dashed lines*). Hip flexion and adduction angles are 0°. Moment arms of the deformed model are similar to the moment arms of the subjects, whose anteversion angles ranged from 44–47°.

We have developed a musculoskeletal model with a "deformable" femur that can compute the moment arms of the hip muscles for a wide range of femoral geometries and joint configurations. In this study, we used this model to estimate the hip rotation moment arms of the medial hamstrings and adductors for the combinations of anteversion angles and body positions exhibited by a representative sample of subjects who walked with excessive internal rotation of the hip. We showed that the semimembranosus, semitendinosus, and gracilis muscles do not have internal rotation moment arms for crouched or internally rotated postures — the body positions commonly assumed by persons with cerebral palsy during walking. This suggests that neither the medial hamstrings nor the gracilis is likely to contribute substantially to excessive internal rotation

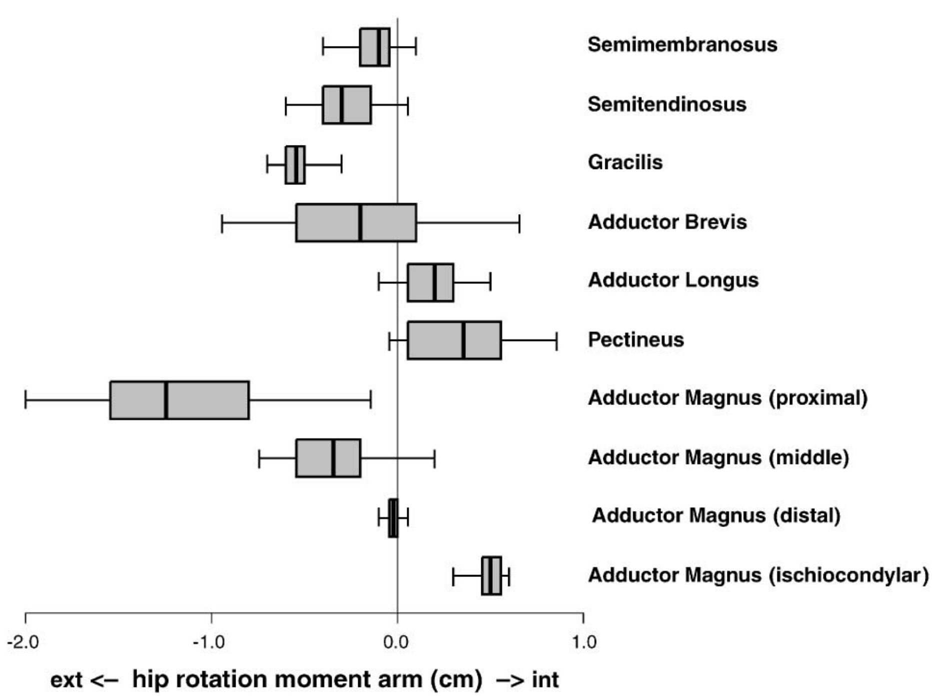

Fig. 7. Peak internal rotation moment arms of the medial hamstrings and adductors during internally rotated gait. Boxplots denote the median (*thick line*), upper and lower quartiles (*shaded region*), and range (*thin lines*) of the peak moment arms estimated with the deformable model for 21 subjects (23 limbs) with cerebral palsy who walked with excessive hip internal rotation.

of the hip. We also determined that the rotational moment arms of the adductor brevis, adductor longus, pectineus, and proximal compartments of the adductor magnus are shifted toward external rotation when the femur is excessively anteverted. Hence, in persons who have anteversion deformities, these muscles are also unlikely to be important contributors to excessive internal rotation of the hip.

It is important to consider some of the limitations of this study. First, we assumed that the hip could be well represented by a ball-and-socket joint. In previous work, we demonstrated that hip flexion and rotation moment arms calculated from MR-based models of three lower limb specimens compared favorably to the moment arms determined experimentally on the specimens (Arnold, 1999; Arnold et al., 2000b). This suggests that our methods for representing hip kinematics in our models and for locating the hip center are adequate — at least for individuals with normal musculoskeletal geometry. However, some persons with cerebral palsy have hips that are subluxed or dislocated. Additional work may be required to develop accurate representations of deformed hips, and to determine how these deformities alter the moment arms of muscles.

Second, we approximated the origins and insertions of muscles as points in our deformable model, even though several of the adductors have broad attachments along the pelvis and femoral shaft. We divided the adductor magnus into four compartments to better characterize this muscle's geometry. Nevertheless, the rotational moment arms of the adductor magnus and other muscles, such as the adductor brevis and adductor longus, may vary slightly with attachment location from the values reported here. We are confident that the trends predicted by our model, however, are representative and accurate.

Third, we developed algorithms to deform the femur of our model after inspecting three-dimensional surface representations of the deformed femurs of only four subjects. These four subjects had anteversion angles that ranged from 34–47°. Some children with cerebral palsy have been reported to have anteversion angles greater than 60° (Ruwe et al., 1992; Laplaza et al., 1993). Imaging of patients with a broader range of deformities in future studies may enable our deformable model to be further tested and improved.

Fourth, for each of our 21 subjects (23 limbs) with internally rotated gait, we deformed the femur of our model based on the subject's assumed anteversion angle (Table 2), which we estimated based on the subject's measured range of hip internal and external rotation (Asymmetry Index, Table 3). Several factors in addition to femoral anteversion angle may influence an individual's hip rotation range of motion, such as degree of acetabular anteversion or laxity of the hip capsule. If we systematically overestimated the femoral anteversion angles of our subjects, then our estimates of the subjects' moment arms could be biased toward external rotation, particularly for the adductors. We elected to use hip rotation range of motion to estimate anteversion angle in this study because it is routinely measured during a patient's clinical exam and is correlated with anteversion of the femur (Staheli et al., 1968). We believe that our Asymmetry Index provided a conservative estimate of most subjects' anteversion angles for several reasons. We limited the maximum anteversion angle predicted by the Index to be 60°, even though some cerebral palsy patients have been reported to have anteversion deformities as large as 80° (e.g., Ruwe et al., 1992). In addition, we designed the Index such that the assumed anteversion angles of our subjects were well distributed between 20 and 60° (Table 2). Only five of the 23 limbs were estimated to have anteversion deformities of 60°; two of the three subjects who were imaged (Subjects A and C) had femoral anteversion angles that were underestimated by our Asymmetry Index. Fortunately, our primary conclusions are not sensitive to our method of estimating anteversion angle; the medial hamstrings and adductors had external rotation moment arms or very small internal rotation moment arms throughout the gait cycle for all of the subjects who were analyzed in this study, including those whose femoral anteversion angles were assumed to be near normal.

Fifth, persons with anteversion deformities frequently undergo derotational osteotomies of the femur, in addition to soft-tissue surgeries, in an effort to improve the rotational alignment of the limb. Alterations in femoral geometry that result from these procedures may change the hip rotation moment arms of muscles. We did not report the effects of such alterations in this study. However, our previous work suggests that changes in the rotational moment arms of the hamstrings and adductors following derotational osteotomy are minimal (Schmidt et al., 1999).

Sixth, we inferred the rotational functions of the medial hamstrings and adductors from analyses of the musculoskeletal geometry. Our deformable model does not include descriptions of the muscle force-generating properties, and our investigation has not considered the potential capacity of the muscles to accelerate the limb in directions opposite to their moment arms through dynamic coupling (Zajac and Gordon, 1989).

Despite these limitations, this study provides important evidence which suggests that the medial hamstrings and adductors cannot generate a substantial hip internal rotation moment in many persons with cerebral palsy who walk with crouched, internally rotated gait. Of the muscles that were analyzed, only the adductor longus, pectineus, and ischiocondylar compartment of the adductor magnus in our model had internal rotation moment arms in some subjects that were greater than 5 mm during walking. The medial hamstrings and the

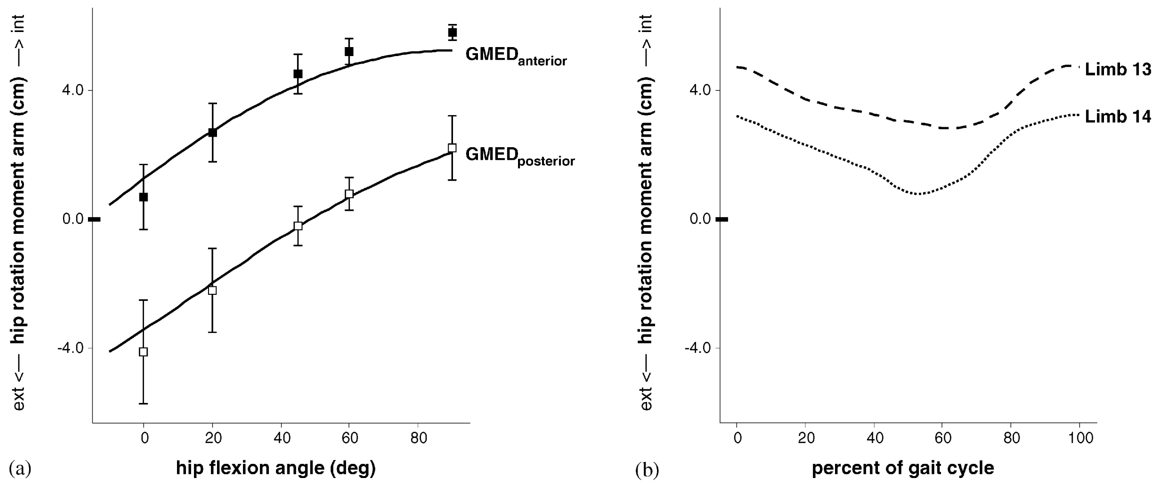

Fig. 8. Hip rotation moment arms of the gluteus medius, plotted vs. hip flexion angle (A) and vs. gait cycle (B). Moment arms calculated with the undeformed model (*solid lines*) compare favorably to the moment arms of the anterior (*filled squares*) and posterior (*open squares*) compartments of the gluteus medius determined experimentally by Delp et al. (mean ± one standard deviation for four specimens, 1999) over a range of hip flexion angles (A). Internal rotation moment arms of the anterior compartment of the gluteus medius, calculated with the deformable model at body positions corresponding to internally rotated gait (B), are approximately four times the moment arms of the medial hamstrings or adductors. Moment arms are shown for two subjects with cerebral palsy and excessive internal rotation of the hip; one subject walked with exaggerated hip flexion (*Limb 13*, *dashed line*), the other walked with relatively normal hip flexion (*Limb 14*, *dotted line*).

other adductors had moment arms that were negligible or *external*, for most subjects, throughout the gait cycle.

To put these data into perspective, we used our deformable model to calculate the hip rotation moment arms of the gluteus medius during crouched, internally rotated gait (Fig. 8). A recent experimental study has shown that the hip rotation moment arms of the gluteus medius and gluteus minimus increase dramatically with hip flexion (Delp et al., 1999). The moment arms of the anterior and posterior compartments of the gluteus medius computed with our model are consistent with this observation (Fig. 8). Since excessive flexion of the hip frequently accompanies internally rotated gait (Bleck, 1987; Gage, 1991), and since the gluteal muscles are typically active and play an important role in walking (Csongradi et al., 1980; Perry, 1992), we hypothesize that the excessive hip flexion, which increases the internal rotation moment arms of the gluteus medius and minimus, is more likely than the hamstrings or adductors to cause internal rotation. The rotational moment arms of the medial hamstrings and adductors in our model were not substantially altered by hip flexion, in contrast to the gluteus medius. This work emphasizes the need to account for altered bone geometry and abnormal joint kinematics when hypothesizing the causes of movement abnormalities and planning surgical treatments.

Acknowledgements

We are grateful to Peter Loan and Ken Smith for help with development of the modeling software; to Silvia Blemker and Deanna Asakawa for their many contributions related to testing of the deformable model; to Stephen Vankoski, Julie Witka, Claudia Kelp-Lenane, and Rob Novak for assistance with data collection; and to Luciano Dias, John Sarwark, and Todd Simmons for discussions we have had regarding causes of excessive hip internal rotation in children with cerebral palsy. This work was supported by NIH Grant R01 HD33929, the United Cerebral Palsy Foundation, and the Powell Foundation.

References

Arnold, A.S., 1999. Quantitative descriptions of musculoskeletal geometry in persons with cerebral palsy: guidelines for the evaluation and treatment of crouch gait. Ph.D. Thesis, Northwestern University, Evanston, IL.

Arnold, A.S., Asakawa, D.J., Delp, S.L., 2000a. Do the hamstrings and adductors contribute to excessive internal rotation of the hip in persons with cerebral palsy? Gait and Posture 11, 181–190.

Arnold, A.S., Komattu, A.V., Delp, S.L., 1997. Internal rotation gait: a compensatory mechanism to restore abduction capacity decreased by bone deformity? Developmental Medicine and Child Neurology 39, 40–44.

Arnold, A.S., Salinas, S., Asakawa, D.J., Delp, S.L., 2000b. Accuracy of muscle moment arms estimated from MRI-based musculoskeletal models of the lower extremity. Computer Aided Surgery 5, 108–119.

Basmajian, J.V., DeLuca, C.J., 1985. Muscles Alive. Williams and Wilkins, Baltimore.

Bleck, E.E., 1987. Orthopaedic Management in Cerebral Palsy. Mac Keith Press, London.

Chong, K.C., Vojnic, C.D., Quanbury, A.O., Letts, R.M., 1978. The assessment of the internal rotation gait in cerebral palsy: an electromyographic gait analysis. Clinical Orthopaedics and Related Research 132, 145–150.

Csongradi, J., Terver, S., Medeiros, J., Bleck, E., 1980. Electromyographic study of the gluteus medius during the gait of normal children and cerebral palsy children. Orthopaedic Transactions 4, 72.

Davis, R.B., Ounpuu, S., Tyburski, D., Gage, J.R., 1991. A gait analysis data collection and reduction technique. Human Movement Science 10, 575–587.

Delp, S.L., Hess, W.E., Hungerford, D.S., Jones, L.C., 1999. Variation of hip rotation moment arms with hip flexion. Journal of Biomechanics 32, 493–501.

Delp, S.L., Loan, J.P., 1995. A graphics-based software system to develop and analyze models of musculoskeletal structures. Computers in Biology and Medicine 25, 21–34.

Delp, S.L., Loan, J.P., Hoy, M.G., Zajac, F.E., Topp, E.L., Rosen, J.M., 1990. An interactive graphics-based model of the lower extremity to study orthopaedic surgical procedures. IEEE Transactions on Biomedical Engineering 37, 757–767.

Dostal, W.F., Soderberg, G.L., Andrews, J.G., 1986. Actions of hip muscles. Physical Therapy 66, 351–361.

Gage, J.R., 1991. Gait Analysis in Cerebral Palsy. Mac Keith Press, London.

Hoffer, M.M., 1986. Management of the hip in cerebral palsy. Journal of Bone and Joint Surgery 68-A, 629–631.

Kadaba, M.P., Ramakrishnan, H.K., Wootten, M.E., 1990. Measurement of lower extremity kinematics during level walking. Journal of Orthopaedic Research 8, 383–392.

Kurosawa, H., Walker, P.S., Abe, S., Garg, A., Hunter, T., 1985. Geometry and motion of the knee for implant and orthotic design. Journal of Biomechanics 18, 487–499.

Laplaza, F.J., Root, L., Tassanawipas, A., Glasser, D.B., 1993. Femoral torsion and neck-shaft angles in cerebral palsy. Journal of Pediatric Orthopaedics 13, 192–199.

Lengsfeld, M., Pressel, T., Stammberger, U., 1997. Lengths and lever arms of hip joint muscles: geometrical analyses using a human multibody model. Gait and Posture 6, 18–26.

Lundy, D.W., Ganey, T.M., Ogden, J.A., Guidera, K.J., 1998. Pathologic morphology of the dislocated proximal femur in children with cerebral palsy. Journal of Pediatric Orthopaedics 18, 528–534.

Mansour, J.M., Pereira, J.M., 1987. Quantitative functional anatomy of the lower limb with application to human gait. Journal of Biomechanics 20, 51–58.

Miller, F., Merlo, M., Liang, Y., Kupcha, P., Jamison, J., Harcke, T., 1993. Femoral version and neck shaft angle. Journal of Pediatric Orthopaedics 13, 382–388.

Murphy, S.B., Simon, S.R., Kijewski, P.K., Wilkinson, R.H., Griscom, N.T., 1987. Femoral anteversion. Journal of Bone and Joint Surgery 69-A, 1169–1176.

Nisell, R., Nemeth, G., Ohlsen, H., 1986. Joint forces in extension of the knee. Acta Orthopaedica Scandinavica 57, 41–46.

Perry, J., 1992. Gait Analysis: Normal and Pathological Function. SLACK, Inc., Thorofare, NJ.

Root, L., 1987. Treatment of hip problems in cerebral palsy. Instructional Course Lectures 36, 237–252.

Ruwe, P.A., Gage, J.R., Ozonoff, M.B., DeLuca, P.A., 1992. Clinical determination of femoral anteversion. Journal of Bone and Joint Surgery 74-A, 820–830.

Schmidt, D.J., Arnold, A.S., Carroll, N.C., Delp, S.L., 1999. Length changes of the hamstrings and adductors resulting from derotational osteotomies of the femur. Journal of Orthopaedic Research 17, 279–285.

Schutte, L.M., Hayden, S.W., Gage, J.R., 1997. Lengths of hamstrings and psoas muscles during crouch gait: effects of femoral anteversion. Journal of Orthopaedic Research 15, 615–621.

Staheli, L.T., Duncan, W.R., Schaefer, E., 1968. Growth alterations in the hemiplegic child. Clinical Orthopaedics and Related Research 60, 205–212.

Sutherland, D.H., Schottstaedt, E.R., Larsen, L.J., Ashley, R.K., Callander, J.N., James, P.M., 1969. Clinical and electromyographic study of seven spastic children with internal rotation gait. Journal of Bone and Joint Surgery 51-A, 1070–1082.

Tachdjian, M.O., 1990. Pediatric Orthopedics. W.B. Saunders, Philadelphia.

Upadhyay, S.S., O'Neil, T., Burwell, R.G., Moulton, A., 1987. A new method using medical ultrasound for measuring femoral anteversion: technique and reliability. Journal of Anatomy 155, 119–132.

van der Helm, F.C.T., Veeger, H.E.J., Pronk, G.M., van der Woude, L.H.V., Rozendal, R.H., 1992. Geometry parameters for musculoskeletal modelling of the shoulder system. Journal of Biomechanics 25, 129–144.

Walker, P.S., Rovick, J.S., Robertson, D.D., 1988. The effects of knee brace hinge design and placement on joint mechanics. Journal of Biomechanics 21, 965–974.

Zajac, F.E., Gordon, M.E., 1989. Determining muscle's force and action in multi-articular movement. Exercise and Sport Sciences Reviews 17, 187–230.