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Evaluation of a Deformable Musculoskeletal Model for Estimating Muscle–Tendon Lengths During Crouch Gait

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Abstract—The hamstrings and psoas muscles are often lengthened surgically in an attempt to correct crouch gait in persons with cerebral palsy. The purpose of this study was to determine if, and under what conditions, medial hamstrings and psoas lengths estimated with a ''deformable'' musculoskeletal model accurately characterize the lengths of the muscles during walking in individuals with crouch gait. Computer models of four subjects with crouch gait were developed from magnetic resonance (MR) images. These models were used in conjunction with the subjects' measured gait kinematics to calculate the muscle–tendon lengths at the body positions corresponding to walking. The lengths calculated with the MR-based models were normalized and were compared to the lengths estimated using a deformable generic model. The deformable model was either left undeformed and unscaled, or was deformed or scaled to more closely approximate the femoral geometry or bone dimensions of each subject. In most cases, differences between the normalized lengths of the medial hamstrings computed with the deformable and MR-based models were less than 5 mm. Differences in the psoas lengths computed with the deformable and MR-based models were also small (<3 mm) when the deformable model was adjusted to represent the femoral geometry of each subject. This work demonstrates that a deformable musculoskeletal model, in combination with a few subject-specific parameters and simple normalization techniques, can provide rapid and accurate estimates of medial hamstrings and psoas lengths in persons with neuromuscular disorders. © *2001 Biomedical Engineering Society.*
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Keywords—Musculoskeletal model, Muscle, Hip, Knee, Gait, Magnetic resonance imaging, Cerebral palsy.

INTRODUCTION

''Tight'' muscles that are thought to restrict movement are often lengthened surgically in an effort to improve walking in persons with cerebral palsy.[5,20] For example, short or spastic hamstrings are presumed to limit knee extension in many children who walk with a troublesome crouch gait; these patients frequently undergo hamstrings lengthening surgery.[20] Excessive flexion of the hip during walking is commonly treated by surgical lengthening of the psoas tendon.[35]

Unfortunately, the outcomes of muscle–tendon surgeries to correct crouch gait and other movement abnormalities in persons with neuromuscular disorders are inconsistent and sometimes unsatisfactory.[20] Lengthening of the hamstrings often decreases excessive knee flexion. However, the hamstrings produce an extension moment about the hip as well as a flexion moment about the knee, and interventions that weaken the hamstrings can lead to other problems during walking, such as exaggerated hip flexion during the stance phase, or insufficient knee flexion and foot clearance during swing.[18,36] Surgical lengthening of the psoas, in some patients, diminishes excessive hip flexion.[35] However, a scientific basis for predicting which patients are likely to benefit from hamstrings and/or psoas lengthening procedures currently does not exist. We believe that analyses of the muscle–tendon lengths during crouch gait may help distinguish patients who have short muscles from those who do not have short muscles, and thus may provide a more effective means to identify candidates who would benefit from surgery.

Several investigators have used computer models of the lower extremity, in conjunction with joint angles measured during gait analysis, to estimate the lengths of the hamstrings and psoas muscles during normal and crouch gait.[16,21,32,37] In these studies, muscle–tendon lengths corresponding to crouch gait were normalized and were compared to the lengths averaged for unimpaired subjects to determine if patients' muscles were operating at normal lengths, or lengths shorter than normal. These analyses have suggested that many individuals with crouch gait do not walk with ''short'' hamstrings; in such cases, factors other than the hamstrings may be contributing to knee flexion.[16,21,32]

Estimates of the muscle–tendon lengths in previous studies were based on a generic model of the lower extremity,[17] representing the musculoskeletal geometry of an average-sized adult male. It is not known how variations in musculoskeletal geometry due to size, age,

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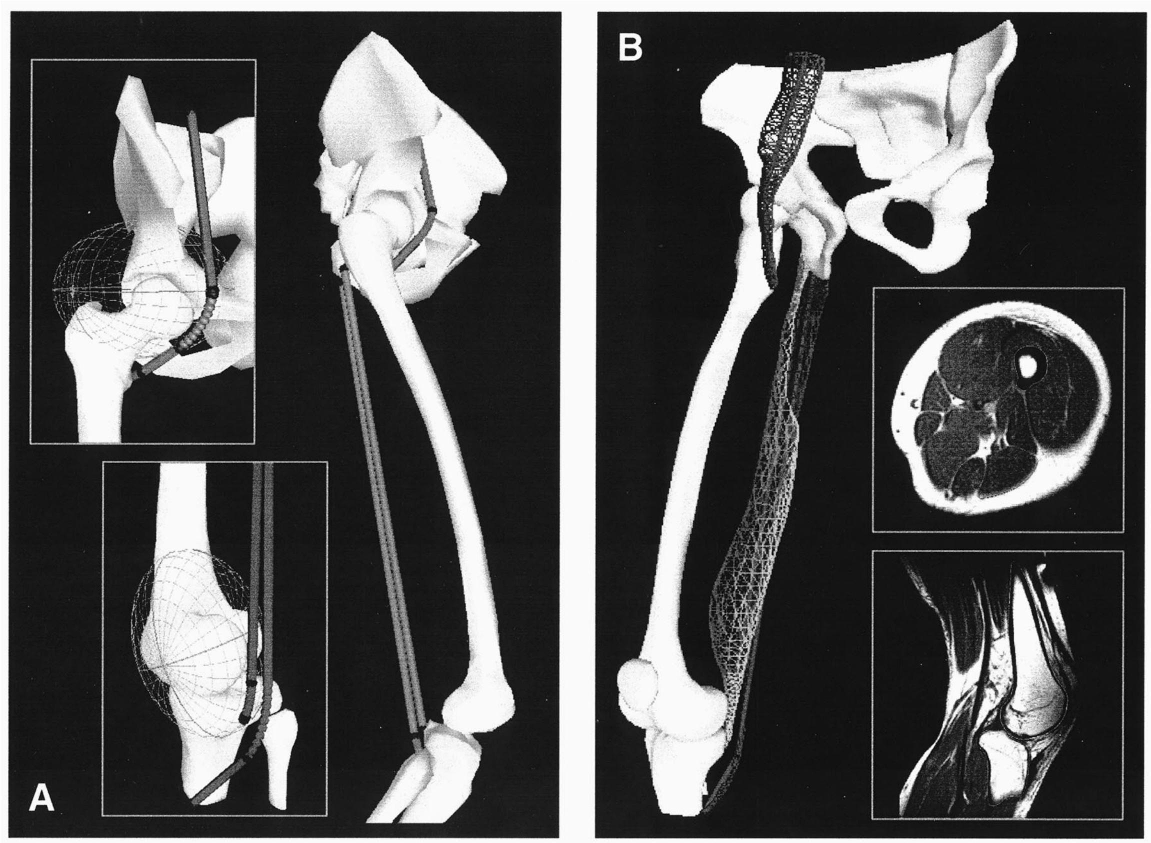

FIGURE 1. Evaluation of a ''deformable'' musculoskeletal model. The normalized lengths of the medial hamstrings and psoas muscles estimated with a deformable generic model (A) were compared to the lengths calculated from models of four individuals with crouch gait developed from MR images [e.g., subject 4 (B)].

or pathology affect the accuracy of muscle–tendon length calculations. In prior studies, the muscle–tendon lengths were normalized by the lengths of the muscles at the anatomical position in an effort to account for differences in size. However, children with cerebral palsy frequently exhibit excessive anteversion of the femur.[5] If this torsional deformity substantially alters the moment arms (i.e., the lever arm, or mechanical advantage of a muscle at a joint) of muscles about the hip, then estimates of the muscle–tendon lengths calculated with a generic model may be inaccurate or misleading. Before generic models can be used to guide treatment decisions for specific patients, the models must be tested.

Schutte *et al.*[32] modified an existing lower limb model[17] to investigate the sensitivity of hamstrings and psoas lengths to femoral anteversion angle. Normalized hamstring lengths computed with the ''deformed'' model were similar to the lengths calculated with the undeformed model; however, normalized psoas lengths varied with deformation of the femur. Schutte *et al.* did not validate their model on the basis of patient-specific descriptions of musculoskeletal anatomy, such as data derived from medical images. Hence, whether a generic model—deformed or undeformed—can provide reliable estimates of the muscle–tendon lengths in persons with femoral deformities remains unclear.

Methods to construct highly accurate, subject-specific models of the musculoskeletal system from magnetic resonance (MR) images have been developed.[1,11,33] At the present time, however, building an MR-based model for every child with crouch gait would be costly and labor intensive. Other investigators have proposed using generic models in combination with multidimensional scaling techniques, or ''hybrid'' models that incorporate just a few subject-specific parameters, to analyze clinical problems.[7,9] However, validation studies that confirm the efficacy of these approaches are lacking. The purpose of this study was to determine if the muscle–tendon lengths estimated with a generic musculoskeletal model and simple normalization techniques are sufficiently accurate to distinguish patients who have short muscles from those who do not have short muscles, or whether subject-specific variations in bone dimensions and/or femoral geometry need to be considered.

METHODS

A graphics-based model of the lower extremity with a ''deformable'' femur was developed, and the accuracy with which this model characterizes the lengths of the medial hamstrings and psoas muscles in individuals with cerebral palsy, at the body positions corresponding to crouch gait, was evaluated (Fig. 1). To test the deformable model, detailed models of four subjects with crouch gait were created from an extensive set of MR images. These models were used, in conjunction with each subject's measured gait kinematics, to determine the lengths of the medial hamstrings and psoas muscles at the joint angles corresponding to walking. The lengths calculated with the MR-based models were normalized and were compared to the lengths estimated using four variations of our deformable generic model. In the first variation, the deformable model was left undeformed and unscaled. In subsequent variations, the deformable model either was deformed or was scaled to more closely approximate the femoral geometry or bone dimensions of each subject.

Development of the Deformable Generic Model

The deformable musculoskeletal model developed for this study 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 psoas muscles for an average-sized adult male. This model is similar to the deformable lower limb models we have used in previous studies,[31] with the following improvements. First, we refined the locations of the muscle attachments reported by Delp *et al.*[17] to be consistent with three-dimensional surface representations of the muscles and bones of three lower extremity cadaveric specimens generated from MR images. Second, we implemented a description of tibiofemoral kinematics that accounts for the three-dimensional rotations and translations of the tibia relative to the femur;[40] in previous models, we neglected the rotations of the tibia in the frontal and transverse planes. Third, we defined ''wrapping surfaces,''[39] in addition to ''via points,''[17] to simulate interactions between the muscles and surrounding anatomical structures, thereby providing an improvement over previous models that used straight-line approximations of the muscle–tendon paths. Finally, 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 subjects with cerebral palsy constructed from MR images. Our resulting model was capable of estimating the lengths of the medial hamstrings and psoas muscles for a range of femoral deformities commonly observed in persons with cerebral palsy and a variety of body positions, including hip and knee angles that corresponded to normal and crouch gait.

We defined the bone geometry, joint kinematics, and muscle–tendon paths of our deformable model using a musculoskeletal modeling package, SIMM.[13] 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,[1] 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.[29,40] Our procedures for establishing the segment coordinate systems and joint kinematics have been reported in detail previously.[1]

The paths of the semimembranosus and semitendinosus muscles, which comprise the medial hamstrings, and the psoas muscle were defined for a range of hip and knee motions. The line of action of each muscle was characterized by a series of line segments. The attachment sites of the muscles were identified, and wrapping surfaces and via points were introduced to simulate underlying structures and other anatomical constraints. We refined 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. Although the psoas originates from the transverse processes of the lumbar vertebrae, we fixed its origin to the model's pelvis reference frame, rather than to a separate sacral or lumbar reference frame. Hence, changes in the length of the psoas in our model reflect changes in hip angles only.

We prescribed the paths of the muscles through a range of hip and knee motions by specifying wrapping surfaces and via points as follows. First, for each of three lower extremity cadaveric specimens, we created a graphics-based kinematic model of the hip joint, the tibiofemoral joint, and the surrounding musculature from MR images.[1] Second, for each muscle, 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 designed the path of each muscle to be consistent with the muscle surfaces constructed from MR images, while minimizing penetration into bones or other muscles. For the medial hamstrings, wrapping surfaces were positioned at the distal femur to prevent the muscle–tendon paths from penetrating the posterior femoral condyles and adjacent soft tissues with knee extension. A via point

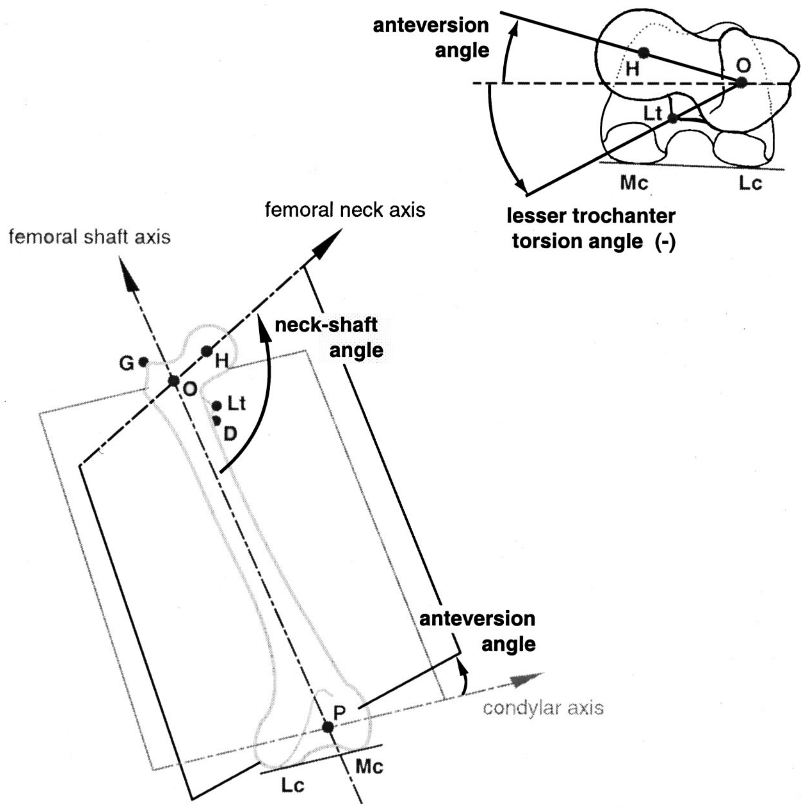

FIGURE 2. Description of femoral geometry: *H* is the center of the femoral head, *G* is the most superior point on the greater trochanter, *D* is the most distal point on the lesser trochanter, *Lt* is the tip of the lesser trochanter, *P* is the attachment of the posterior cruciate ligament, *O* is 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 passing 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*. Neck–shaft angle is the angle formed by the femoral neck axis and the femoral shaft axis. Lesser trochanter torsion angle is the angle formed by the plane of the condylar axis and the plane, which passes through points *O*, *P*, and *Lt*. If point *Lt* is anterior to the condylar axis, this angle is defined as positive. If point *Lt* is posterior to the condylar axis, the angle is defined as negative. The figure is adapted from Murphy *et al.* (Ref. 27) and Calais-Germain (Ref. 8).

was added proximal to the insertion of the semitendinosus to mimic the constraints produced by surrounding connective tissues. For the psoas, a wrapping surface was placed near the acetabulum to characterize the wrapping and sliding of the muscle over the pelvic brim and hip capsule. A via point representing the ''effective'' origin of the psoas was fixed at the pelvic brim. Another via point was located proximal to the muscle's insertion to prevent the muscle–tendon path from penetrating the femoral neck with hip internal rotation. We verified the efficacy of each algorithm by comparing the muscle moment arms calculated with the MR-based models of the three cadaveric specimens to the moment arms determined experimentally on the same specimens.[1] Once an algorithm was developed that could predict the muscle moment arms with sufficient accuracy (i.e., moment arms within 10% of the experimental data) for all three specimens, the same algorithm was used to specify the muscle–tendon paths of our deformable generic model.

We developed techniques to deform the femur of our generic model to represent excessive anteversion and other deformities commonly observed in persons with crouch gait. To do this, we compared the undeformed femur of our generic model to three-dimensional surface representations of the deformed femurs of four subjects with cerebral palsy generated from MR images. We hypothesized that the deformed femurs of the subjects could be well characterized by three geometric parameters—anteversion angle, neck-shaft angle, and lesser trochanter torsion angle—and we developed a mathematical description of each parameter (Fig. 2). The subjects with cerebral palsy who were imaged in this study had femoral anteversion angles that ranged from 34° to 47°, neck-shaft angles that ranged from 129° to

TABLE 1. Characteristics of the cerebral palsy subjects and the undeformed generic model.

	Subject 1	Subject 2	Subject 3	Subject 4	Generic model
Gender	F	M	M	M	M
Age (yrs)	7	14	14	27	adult
Height (cm)	126	132	169	165	NA[e]
Weight (kg)	24.7	25.6	51.9	45.4	NA[e]
Femur length[a]	31.1	36.0	40.3	37.5	39.6
Anteversion angle[b]	47	34	44	46	20
Neck-shaft angle[b]	129	131	138	142	125
Lesser trochanter torsion angle[b]	−16	−14	−7	+13	−33
Hip flexion during stance phase[c] of gait (max/min)	51/3	14/7	39/17	61/28	NA[e]
Knee flexion during stance phase[d] of gait (max/min)	33/3	39/29	39/26	83/73	NA[e]

[a]Superior−inferior dimension from center of femoral head to midpoint between femoral epicondyles, in units of cm.

[b]Defined in Fig. 2, in units of degrees.

[c]Angle formed in the sagittal plane (i.e., the plane perpendicular to the medial−lateral axis of the pelvis, as defined by the left and right anterior superior iliac spines) between the long axis of the thigh and a vector 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.

[d]Angle formed in the sagittal plane (i.e., the plane perpendicular to the medial−lateral axis of the femur, as defined by a knee alignment device) 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.

[e]Not applicable.

142°, and lesser trochanter torsion angles that varied from −16° to +13° (Table 1). The femur of our undeformed generic model has an anteversion angle of 20° and a neck-shaft angle of 125°, which are within the normal range for unimpaired adults.[10,38] Our definitions of femoral anteversion angle and neck-shaft angle are consistent with descriptions that have been used in the past by clinicians and other investigators.[27] A definition of lesser trochanter torsion angle, to our knowledge, has not appeared previously in the literature. We examined the orientation of the lesser trochanter carefully in this study because it influences the path of the psoas, and because it varied substantially among our four subjects.

We altered the femoral anteversion angle, neck-shaft angle, and lesser trochanter torsion angle of our deformable model by rotating and/or translating the bone vertices that make up the femoral head, neck, and shaft. To increase the anteversion angle, the femoral head and neck were rotated 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 (Fig. 2). To increase the neck-shaft angle, the femoral head and neck were rotated superiorly about an axis through the diaphysis of the femur, perpendicular to the plane formed by the femoral neck and shaft. To adjust the lesser trochanter torsion angle, the lesser trochanter was rotated anteriorly or posteriorly about the femoral shaft axis. After each transformation, the bone vertices proximal to the femoral condyles were translated as needed to restore the position of the femoral head in the acetabulum. The insertion of the psoas on the lesser trochanter was displaced with the bone vertices. Hence, the length of the psoas at the anatomical position, the moment arms, and the length changes of the muscle during movement were altered by these deformities. The position of the knee center with respect to the hip center in our deformable model was not changed; thus, the paths of the medial hamstrings were not affected.

Construction of the MR-Based Models

We assessed the accuracy with which our deformable model could estimate medial hamstrings and psoas lengths during crouch gait by creating detailed, MR-based models of four subjects selected from the cerebral palsy clinics at the Children's Memorial Medical Center in Chicago. Each subject underwent gait analysis using a five-camera motion measurement system (VICON, Oxford Metrics, Oxford, U.K.). The subject's three-dimensional gait kinematics were computed as described by Kadaba *et al.*,[22] based on estimates of the joint center locations as suggested by Davis *et al.*[12] The limb that showed the greatest degree of knee flexion was selected for further analysis. The subjects ranged in age from 7 to 27 yr and walked with different gait abnormalities, ranging from a relatively mild crouch gait to a severe crouch gait (Table 1). None of the subjects had undergone previous surgery, and all were able to walk without orthoses or other assistance. All subjects and/or their parents provided informed written consent.

The process of creating each MR-based model consisted of six steps. Step 1 was to acquire the MR images,

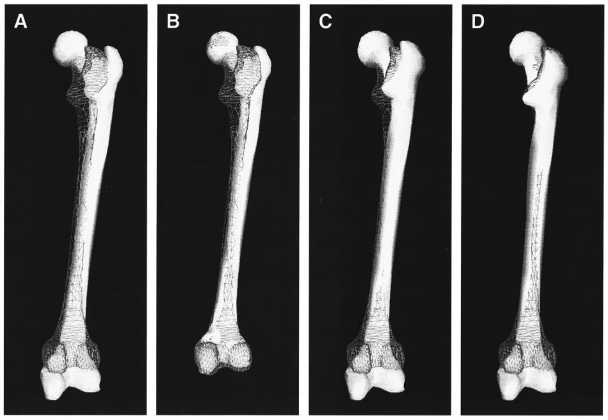

FIGURE 3. Deformation and scaling of the generic model. The undeformed femur of the generic model [(A), solid bone] was altered to more closely approximate the bone dimensions and femoral geometry of each subject (e.g., subject 4, wireframe bone) by scaling the model along anatomical axes (B), increasing its femoral anteversion angle (C), or adjusting its femoral anteversion angle, neck-shaft angle, and lesser trochanter torsion angle (D).

TABLE 2. Factors[a] for scaling the generic model to the MR-based models.

	Subject 1	Subject 2	Subject 3	Subject 4
Pelvis				
AP dimension[b]	0.65	0.61	0.95	0.90
SI dimension[c]	0.82	0.72	0.98	0.88
ML dimension[d]	0.69	0.64	1.01	0.92
Femur and Tibia				
AP dimension[e]	0.74	0.79	0.86	0.85
SI dimension[f]	0.78	0.89	1.01	0.93
ML dimension[g]	0.86	0.89	1.03	0.94

[a]Scale factors represent the ratio of the MR-based model dimension to the generic model dimension.
[b]Anterior–posterior dimension from anterior superior iliac spine (ASIS) to hip center.
[c]Superior–interior dimension from ASIS to ischial tuberosity.
[d]Medial–lateral dimension from right ASIS to left ASIS.
[e]Length of the tibial plateau in the sagittal plane.
[f]Distance from hip center to the midpoint between spheres fit to the medial and lateral posterior femoral condyles.
[g]Medial–lateral dimension between femoral epicondyles.

which was done using a 1.5 T Signa MR Scanner (GE Medical Systems, Milwaukee, WI). Approximately 200 T1-weighted spin echo images were collected for each subject. Step 2 was to identify and outline the anatomical structures of interest on each image. These structures included the pelvis, sacrum, femur, tibia, semimembranosus, semitendinosus, and psoas. Step 3 was to generate three-dimensional surface reconstructions of each structure from the two-dimensional outlines, and step 4 was to register the surfaces from adjacent series of images. This yielded an accurate representation of each subject's anatomy at one limb position. Step 5 was to define kinematic models of the hip and tibiofemoral joints based on each subject's bone surface geometry. Step 6 was to characterize the muscle–tendon paths, as described above, for a range of hip and knee motions. Our protocol for MR imaging, our techniques for surface reconstruction and registration, and our methods for specifying the joint kinematics are described in detail elsewhere.[1]

Comparison of Lengths Calculated with the Deformable and MR-Based Models

Muscle–tendon lengths determined from the MR-based models were used to examine the accuracy of the lengths estimated with the deformable generic model. For each of our four cerebral palsy subjects, the lengths of the semimembranosus, semitendinosus, and psoas muscles were calculated at the limb positions corresponding to the subject's measured gait kinematics. Semimembranosus length was calculated between the muscle's origin and insertion. Semitendinosus length was calculated between the muscle's origin and its effective insertion near the posterior femoral condyles. Psoas length was computed between the muscle's insertion and its effective origin at the pelvic brim; hence, we assumed that changes in psoas length due to rotations at the lower lumbar spine and lumbosacral joint were negligible.

The lengths of the muscles during crouch gait L_i were normalized based on the maximum averaged length $L_{\max}$ and the minimum averaged length $L_{\min}$ of the muscle during normal gait as follows:

$$\hat{L}_i = (L_i - L_{\min})/(L_{\max} - L_{\min}),$$

where $\hat{L}_i$ is the normalized length of the muscle at the ith point of the gait cycle. Values of $L_{\max}$ and $L_{\min}$ were obtained for each model based on the measured gait kinematics of 18 unimpaired subjects. This normalization technique is relevant because the muscle–tendon lengths of cerebral palsy subjects are often compared to averaged data from unimpaired subjects to determine if a muscle is shorter or longer than normal during walking.[16,32,37] Using this technique, the normalized lengths of the muscles for unimpaired subjects, during normal walking, were similar when calculated with the different models.

The muscle–tendon lengths calculated with each subject's MR-based model were normalized and were compared to the lengths estimated using four variations of the deformable generic model: (i) the undeformed generic model, (ii) the undeformed generic model scaled to the subject along anatomical axes, (iii) the generic model deformed to match the subject's femoral anteversion angle, called Deformed Model A, and (iv) the generic model deformed to match the subject's anteversion angle, neck-shaft angle, and lesser trochanter torsion angle, called Deformed Model B (Fig. 3). In variation II, we altered the bone dimensions (change in size) prior to normalization; in variations III and IV, we introduced localized changes in the geometry of the proximal femur (change in shape) prior to normalization. Our goal was to determine whether the normalized muscle–tendon lengths estimated with our undeformed, unscaled generic model are generally of sufficient accuracy to guide the planning of muscle–tendon surgery, or whether the femoral geometry or bone dimensions of patients need to be considered.

In variation II, the undeformed generic model was scaled to each subject along anterior–posterior, medial–lateral, and superior–inferior axes using a linear homogeneous transformation.[24] All bones, joints, muscle attachments, via points, and wrapping surfaces in the model were scaled. Different scale factors were applied to structures associated with the pelvis, and to structures associated with the femur and tibia. We chose dimensions for computing the scale factors according to two criteria. First, we required each dimension to be based on skeletal landmarks that could be palpated or reasonably estimated. This ensured that the scaling scheme would be applicable to other individuals with crouch gait, if desired, without having to build a model of every patient from image data. Second, we attempted to scale the bones of the generic model to the bones of each subject as accurately as possible. We used an iterative closest point method[4] and a Gauss–Newton nonlinear least-squares algorithm (MATLAB Optimization Toolbox, The MathWorks, Natick, MA) to calculate scale factors that minimized the total distance between the bone vertices in the generic model and bone vertices in each MR-based model. We then selected anatomical dimensions for scaling that produced scale factors similar to the optimization solution (Table 2).

The normalized lengths of the medial hamstrings and psoas muscles were plotted at every 2% of the gait cycle. For each subject and each model, the peak muscle–tendon lengths during crouch gait were computed. We were particularly interested in the accuracy with which our deformable generic model could estimate the peak lengths of the muscles during crouch gait, because these are the times in the gait cycle when tight muscles may restrict movement. Differences in the peak lengths calculated with the MR-based models and estimated with each version of the deformable model were expressed in ''standard deviations'' (SD) of the peak lengths during normal gait, determined from the averaged gait kinematics of 18 unimpaired subjects. This unit enabled consistent comparisons of the errors to be made across models. For the semimembranosus, the equivalent length of one SD ranged from 2.8 to 4.4 mm, as calculated with the different models. For the semitendinosus, the equivalent length of one SD ranged from 3.4 to 5.4 mm. For the psoas, the equivalent length of one SD ranged from 1.8 to 3.0 mm.

RESULTS

The peak medial hamstrings lengths computed with the undeformed generic model differed by at most 1 SD from the peak lengths calculated with each MR-based model (Table 3), with the exception of the semitendinosus of subject 4 (Fig. 4). The discrepancy for subject 4 reflects the abnormally posterior path of the subject's semitendinosus in the popliteal region, which was evident in the MR images. Scaling the generic model to each subject along anatomical axes prior to normalization did not improve the accuracy of semimembranosus or semitendinosus lengths estimated with the model (Table 3). The normalized lengths of the medial hamstrings estimated with the generic model, scaled or unscaled, were not systematically greater or smaller than the lengths calculated with the MR-based models. Also, errors in the normalized muscle–tendon lengths during crouch gait were not consistently increased or decreased at any particular part of the gait cycle.

Errors in the peak psoas lengths computed with the undeformed generic model during crouch gait ranged from 0.5 to 1.8 SD (Table 3). The smallest error was obtained for subject 2, who was the least impaired and least deformed subject (Fig. 5). For the other three subjects, the undeformed model underestimated the normalized length of the psoas throughout the gait cycle. Scaling the undeformed model to each subject along

TABLE 3. Errors[a,b] in peak muscle–tendon lengths estimated with the deformable model.

	Subject 1	Subject 2	Subject 3	Subject 4
Semimembranosus[c]				
Generic model	−0.9	−0.2	+1.0	+0.6
Scaled model	−1.0	−1.0	+1.2	+0.7
Semitendinosus[d]				
Generic model	−0.9	+0.3	+0.3	+3.2
Scaled model	−1.0	−0.5	+0.6	+3.3
Psoas[e]				
Generic model	−0.8	+0.5	−1.8	−1.6
Scaled model	−0.7	+0.6	−1.8	−1.8
Deformed Model A	+0.3	0.0	−0.9	−0.9
Deformed Model B	+0.2	0.0	−0.2	−0.6

[a]Error defined as the difference in peak muscle–tendon length during crouch gait calculated with the deformable and MR-based models, expressed in standard deviations of the peak length during normal walking.

[b]Positive value indicates that the peak length estimated with the deformable model is greater than the peak length computed with the MR-based model.

[c]1 SD=2.8–4.4 mm, as calculated with the different models.

[d]1 SD=3.4–5.4 mm, as calculated with the different models.

[e]1 SD=1.8–3.0 mm, as calculated with the different models.

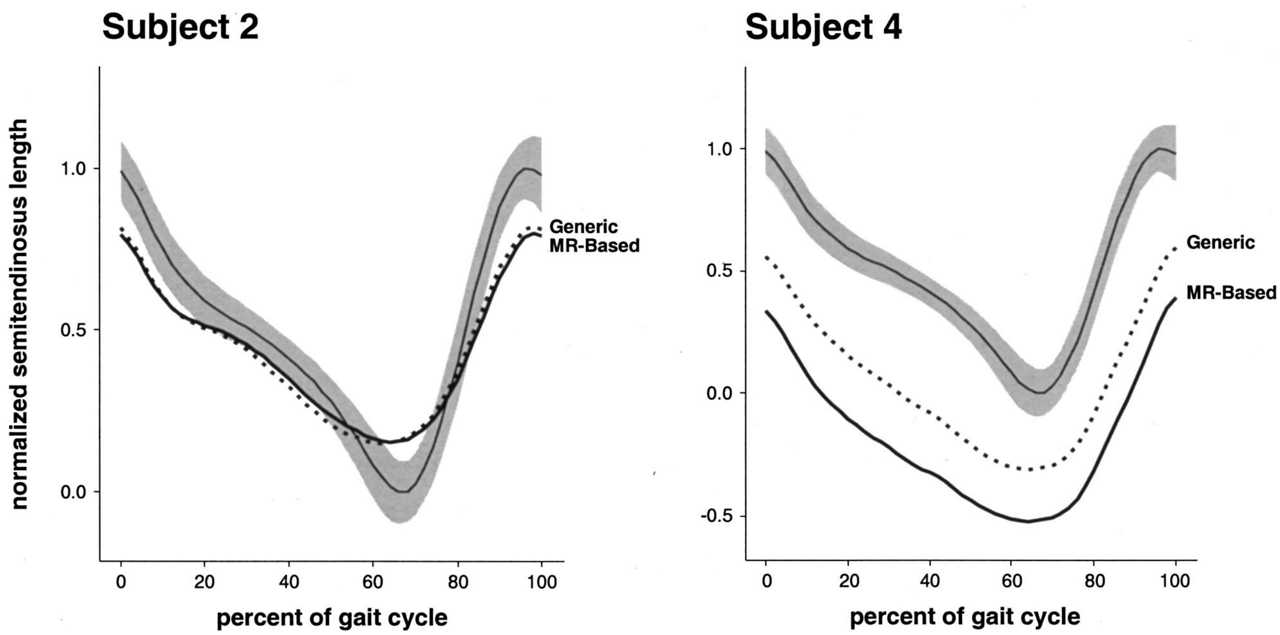

FIGURE 4. Plots of normalized semitendinosus length vs gait cycle, estimated with the undeformed generic model (dotted line) and calculated with the MR-based model (solid line) for subject 2 (best result) and subject 4 (worst result). The normalized length of the semitendinosus during normal gait, averaged for 18 unimpaired subjects (mean±1 SD, shaded region) is shown for comparison.

anatomical axes did not improve the accuracy of the normalized psoas lengths estimated with the model (Table 3).

The generic model more accurately estimated the length of the psoas during crouch gait when subject-specific variations in femoral geometry were considered (Fig. 5). Errors in the peak psoas lengths computed for Deformed Model A ranged in magnitude from 0 to 0.9

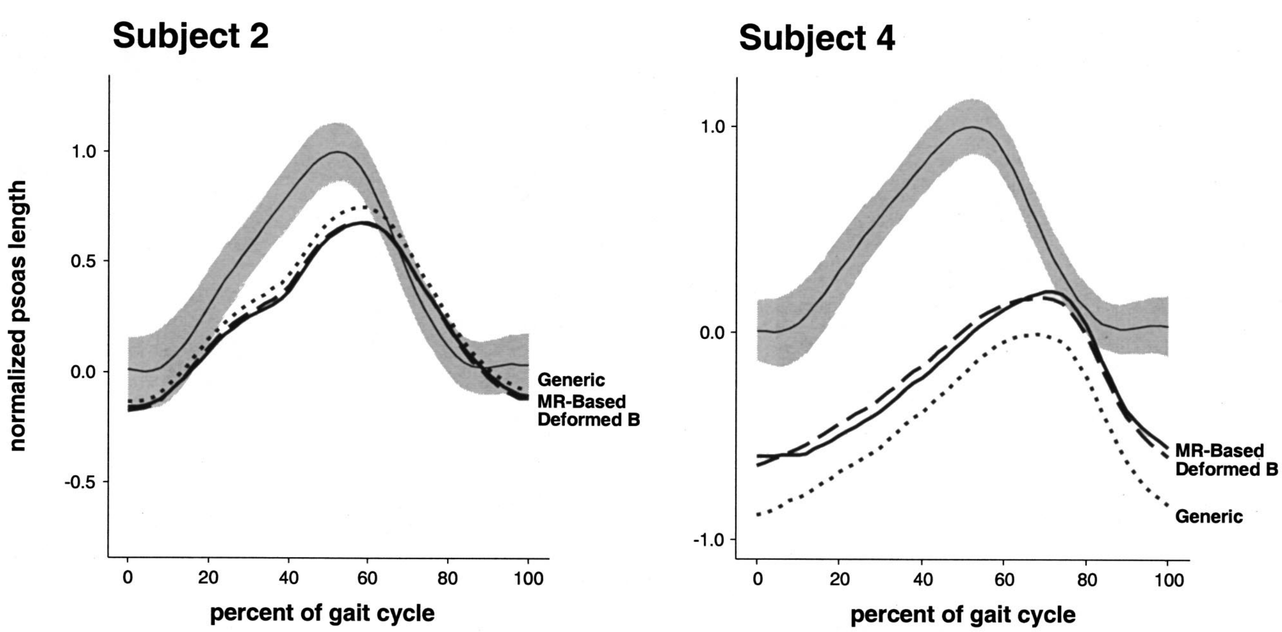

FIGURE 5. Plots of normalized psoas length vs gait cycle, estimated with the undeformed generic model (dotted line), the generic model deformed to match the subject's femoral anteversion angle, neck–shaft angle, and lesser trochanter torsion angle (Deformed Model B, dashed line), and calculated with the MR-based model (solid line) for subject 2 (best result) and subject 4 (worst result). The normalized length of the psoas during normal gait, averaged for 18 unimpaired subjects (mean±1 SD shaded region), is shown for comparison.

SD; errors computed for Deformed Model B ranged from 0 to 0.6 SD (Table 3).

DISCUSSION

Biomechanical models that compute the lengths and moment arms of soft tissues with sufficient accuracy have tremendous potential to impact the design, planning, and evaluation of a variety of musculoskeletal procedures. Surgeons frequently introduce changes in muscle force- and moment-generating capacities by modifying the lengths or moment arms of muscles. Predicting the biomechanical consequences of surgical alterations, therefore, requires detailed knowledge of muscle–tendon lengths and moment arms before and after surgery. Generic models, representing normal adult musculoskeletal geometry, have been used to simulate tendon lengthenings,[15] tendon transfers,[7,14,25] osteotomies,[3,6,19,31] and other procedures. These analyses have determined how variations in surgical parameters affect muscle–tendon lengths, moment arms, force-generating capacities, and joint contact forces postoperatively—data that are relevant to surgical planning. However, no study has reported how variations in musculoskeletal geometry across patients might influence the simulation results. We believe that the accuracy with which musculoskeletal models represent individuals of different sizes, ages, and pathologies must be investigated before simulations can be widely used to guide treatment decisions for patients.

Descriptions of muscle–tendon lengths are particularly applicable to the planning of interventions for crouch gait and other movement abnormalities because a tight muscle that restricts movement is often lengthened surgically. In this study, we developed models of four individuals with crouch gait from MR images, and we used these models to examine the accuracy of medial hamstrings and psoas lengths estimated with a deformable generic model. In seven of eight cases, differences in the normalized lengths of the semimembranosus and semitendinosus muscles estimated with the deformable model and calculated with the MR-based models were less than 5 mm, or about 1 SD of the lengths averaged for unimpaired subjects during normal gait. Errors in the normalized psoas lengths estimated with the deformable model were also less than 1 SD of the averaged lengths for unimpaired subjects—if the model was appropriately deformed to approximate the femoral geometry of each subject.

To put these errors into perspective, we calculated the length changes of the medial hamstrings for a 30° decrease in popliteal angle, a typical improvement that might result from hamstrings lengthening surgery. Popliteal angle measures the degree to which the knee can be passively extended with the hip flexed 90°.[5,23] Several studies have reported average popliteal angles near 60° before surgery and average popliteal angles near 30° following surgical lengthening of the hamstrings in persons with cerebral palsy.[2,18] We determined, using our deformable model, that a decrease in popliteal angle from 60° to 30° increases the lengths of the medial hamstrings by about 2.5 cm. This is approximately five times larger than errors in the muscle–tendon lengths during crouch gait estimated with our deformable model. Based on these data, we believe that a deformable model, in conjunction with a few subject-specific parameters and simple normalization techniques, can provide reasonable estimates of the muscle–tendon lengths in most cases. Whether such estimates can aid surgical decision-making for persons with cerebral palsy is a focus of our ongoing work.

We found that scaling our undeformed generic model to each subject along orthogonal axes, using scale factors based on the bone dimensions, did not reduce errors in the normalized muscle–tendon lengths estimated with the undeformed, unscaled model. This result implies that our scheme for normalizing the muscle–tendon lengths was effective in minimizing errors that otherwise would have been caused by size variations between the generic model and each MR-based model. The data also suggest that our homogeneous scaling method was not helpful in reducing discrepancies in the muscle–tendon lengths from other potential sources, such as nonsystematic errors caused by variations in the muscle attachment locations relative to the joint centers. In a study of elbow muscles in ten upper extremity specimens, Murray *et al.*[28] reported that the dimensions of the humerus, ulna, and radius bones were not good predictors of elbow flexion moment arms unless the bone dimensions were also correlated with the shortest distances between the muscle attachments and the axis of elbow flexion. The fact that our simple scaling method did not enhance the accuracy of the normalized muscle–tendon lengths estimated with the generic model is consistent with Murray *et al.*'s observations.

A large difference was observed in the normalized length of the semitendinosus estimated with the generic model and calculated with the MR-based model of subject 4 (Fig. 4 and Table 3), due to the abnormally posterior path of the subject's semitendinosus tendon relative to the knee. Whether any generic model and/or normalization scheme would accurately characterize the semitendinosus length of subject 4 is questionable, and the incidence of such abnormalities among persons with crouch gait is not known. However, scaling algorithms based on the muscle's effective attachments could perhaps reduce such errors in the muscle–tendon lengths estimated with a generic model. Developing a practical method to locate the muscle attachments relative to the hip and knee joints in persons with cerebral palsy poses a challenge, but minimal MR protocols, or three-dimensional ultrasound techniques, might be feasible.

The normalized lengths of the psoas estimated with the undeformed generic model were shorter than the lengths calculated with the MR-based models for three of the four subjects in this study. It is likely that differences in femoral anteversion angle, neck-shaft angle, lesser trochanter torsion angle, and neck length all contributed to variations in the muscle–tendon lengths across models. Differences in size and development of the lesser trochanter, for the younger subjects, were also factors. All of the subjects in this study walked with psoas muscles that were substantially shorter than normal; hence, none would have been ''misclassified'' as having a psoas of normal length based on the predictions of the undeformed model. However, the tendency of the undeformed model to underestimate psoas length in persons with femoral deformities, and in particular, the potential for the model to underestimate psoas length in patients who may not have a short psoas, is cause for concern. This observation agrees with the conclusions of Schutte *et al.*[32]

Reasonably accurate estimates of psoas length were obtained for all four subjects in this study when the femoral anteversion angle of the generic model was altered to match the anteversion angle of each subject. Thus, for future analyses of psoas lengths in patients with femoral deformities, use of a deformable model is recommended. The femoral anteversion angle of a patient could be rapidly estimated from ultrasound images,[26,38] palpation of the greater trochanter,[30] or measurement of the patient's hip rotation range of motion.[5] Such methods for determining anteversion angle may not be as accurate as the methods used in this study; however, the resulting muscle–tendon lengths are likely to be more accurate than would be obtained from an undeformed generic model.

It is important to keep in mind some of the limitations of this study. First, we assumed that the MR-based models provided accurate estimates of the muscle–tendon lengths in the subjects with cerebral palsy. Our algorithms to define the muscle–tendon paths were initially used to construct models of three lower extremity cadaveric specimens; hence, they were validated through careful anatomical dissections and detailed comparisons of the muscle moment arms calculated with the models to the moment arms determined experimentally on the same specimens.[1] Nevertheless, errors in the joint kinematics or invalid assumptions about how the muscle–tendon paths change with bone deformities could have produced errors in the muscle–tendon lengths determined from the MR-based models. For example, we assumed that the hip could be well represented by a ball-and-socket joint, even though some persons with cerebral palsy have hips that are subluxed or dislocated. Subjects 3 and 4, in fact, showed some evidence of hip subluxation. In future studies, detailed analyses of how skeletal orientations and muscle–tendon paths change with joint configuration *in vivo* may improve the reliability of kinematic models derived from static MR images.

Second, we developed models from MR images of only four subjects with crouch gait. The four individuals who were imaged spanned a wide range of sizes and ages; their femur dimensions from hip center to knee center ranged from 31 to 40 cm, and their ages ranged from 7 to 27 yr. They also exhibited various musculoskeletal impairments, with femoral anteversion angles ranging from 34° to 47° and gait abnormalities ranging from a ''jump knee'' pattern[34] to a severe crouch. Nevertheless, whether our deformable model is suitably accurate for estimating medial hamstrings or psoas lengths in individuals with more severe bone deformities, or in patients with gait patterns much different from the subjects who were analyzed, remains untested.

We calculated the length of the psoas in this study from the muscle's effective origin at the pelvic brim to its insertion on the lesser trochanter. Thus, our model did not account for changes in psoas length due to rotations at the lower lumbar spine and lumbosacral joint. Some children with crouch gait, however, exhibit increased lumbar lordosis in addition to excessive hip flexion. The degree to which these variations in spine position affect the length of the psoas is unknown. If this issue is to be addressed in future studies, a system to accurately measure the kinematics of the lumbar spine during crouch gait is needed.

Finally, accurate estimates of hamstrings and psoas lengths during crouch gait may be insufficient to determine the most appropriate treatment. Ideally, recommendations for muscle–tendon surgery might be based on quantitative descriptions of how a procedure is likely to alter the muscle force-generating properties, and on knowledge of how the altered muscle force-generating properties are likely to influence a patient's gait. Analysis of the muscle–tendon lengths only weakly approximates this ideal. Such analyses can determine if a muscle is ''short'' during crouch gait, but such analyses cannot currently explain *why* a muscle is short. Furthermore, several factors other than tight hamstrings or psoas muscles may contribute to crouch gait such as: weak hip extensors, deficient plantar flexors, or problems with balance. Certainly, much more work is needed to understand how surgical lengthening of the hamstrings and psoas muscles affect the muscle actions, and to determine how these muscles, altered by pathology or surgery, contribute to the motions of the limb segments during crouch gait.

Despite these limitations, we remain cautiously optimistic that analyses of hamstrings and psoas lengths during crouch gait, based on a well-tested deformable model, could aid in the development of more effective treatment plans. Surgical recommendations for persons with crouch gait, at present, are based on qualitative observations of the patient's gait, assessment of variables measured during clinical examination and gait analysis, and the intuition and experience of the clinical team. Muscle–tendon lengths provide information that is not readily available from gait analysis or clinical examination, but which is relevant to surgical planning. The deformable model presented here enables rapid and accurate estimation of hamstrings and psoas lengths for individuals with a range of movement abnormalities and femoral deformities.

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