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Do the hamstrings and adductors contribute to excessive internal rotation of the hip in persons with cerebral palsy?

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Abstract

Children with cerebral palsy frequently walk with excessive internal rotation of the hip. Spastic medial hamstrings or adductors are presumed to contribute to the excessive internal rotation in some patients; however, the capacity of these muscles to produce internal rotation during walking in individuals with cerebral palsy 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 who walk with a crouched, internally-rotated gait. Highly accurate computer models of three subjects with cerebral palsy were created from magnetic resonance images. These subject-specific models were used in conjunction with joint kinematics obtained from gait analysis to calculate the rotational moment arms of the muscles at body positions corresponding to each subject's internally-rotated gait. Analysis of the models revealed that the medial hamstrings, adductor brevis, and gracilis had negligible or *external* rotation moment arms throughout the gait cycle in all three subjects. The adductor longus had an internal rotation moment arm in two of the subjects, but the moment arm was small (< 4 mm) in each case. These findings indicate that neither the medial hamstrings nor the adductor brevis, adductor longus, or gracilis are likely to be important contributors to excessive internal rotation of the hip. This suggests that these muscles should not be lengthened to treat excessive internal rotation of the hip and that other factors are more likely to cause internally-rotated gait in these patients.

Keywords: Cerebral palsy; Hip; Gait; Musculoskeletal model; Moment arm

1. Introduction

Excessive internal rotation of the hip is a common, troublesome gait abnormality among persons with cerebral palsy. Spastic medial hamstrings or adductors are often considered to be one of the factors that contributes to the excessive internal rotation based on evidence from electromyographic recordings [1,2]. Surgical lengthening of these muscles, performed in isolation or in combination with other procedures, is frequently expected to decrease the abnormal internal rotation of the hip and improve the alignment of the limb during walking [3–5].

Several investigators have reported hip rotation moment arms of the hamstrings and adductor muscles [6–8]. These studies — based on adult-sized cadavers and computer models that represent normal musculoskeletal geometry — suggest that the medial hamstrings, adductor brevis, and adductor longus have small (1 cm or less) internal rotation moment arms in the upright, standing position. These data provide insight into the rotational capacity of the muscles for a limited set of body positions; however, it is unclear whether studies performed on adult-sized specimens with normal musculoskeletal geometry accurately characterize the actions of muscles in children with cerebral palsy.

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

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by excessive anteversion of the femur [9], a torsional deformity which 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.

No study has reported rotational moment arms of the hip muscles for children with anteversion deformities who walk with a crouched, internally-rotated gait. As a result, the capacity of the hamstrings and adductors to produce internal rotation in these patients is unknown, and a scientific rationale for lengthening these muscles to treat excessive internal rotation of the hip does not exist.

The goal of this study was to determine if the medial hamstrings or adductors could potentially cause excessive hip internal rotation in persons with cerebral palsy, and if so, to determine which of the muscles were most likely to contribute. From magnetic resonance (MR) images, we developed graphics-based computer models that represent the geometry of the pelvis, femur, and proximal tibia, the kinematics of the hip and the knee, and the paths of the hamstrings and adductor muscles for three subjects with cerebral palsy. The model of each subject was used in conjunction with joint angles measured during gait analysis to examine the rotational moment arms of the muscles at the body positions corresponding to the subject's internally-rotated gait. This paper provides the first quantitative descriptions of hip rotation moment arms in persons with cerebral palsy. The results focus on the semimembranosus, semitendinosus, adductor brevis, adductor longus, and gracilis because these muscles are commonly lengthened in persons who walk with excessive internal rotation of the hip.

2. Methods

Graphics-based musculoskeletal models were developed for three subjects with excessive internal rotation of the hip 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, UK). The limb that showed the greatest degree of hip internal rotation was selected for analysis. The subjects ranged in age from 7 to 27 years and exhibited different gait abnormalities and musculoskeletal impairments (Table 1 and Fig. 1). Subject 1, a 7-year-old female with spastic diplegia, walked with excessive hip flexion, a jump-knee pattern [10], and about 35° of hip internal rotation. Subject 2, a 14-year-old male with spastic hemiplegia, Winters-Gage Type IV [11], walked with about 40° of hip internal rotation on his involved side, some increased hip flexion, and persistent knee flexion. Subject 3, a 27-year-old male with spastic diplegia, walked with about 30° of hip internal rotation and a severe crouch gait. The femoral anteversion angles of the subjects were 47, 44 and 46°, respectively, and were calculated from three-dimensional representations of the bones constructed from MR images. None of the subjects had undergone previous surgery, and all were able to ambulate without orthoses or other assistance. All subjects and/or their parents provided informed consent.

The process of creating each subject-specific model consisted of six steps (Fig. 2). Step 1 was to acquire the MR images, which was done with the GE Signa MRI Scanner (1.5 T, GE Medical Systems, Milwaukee, WI) at the Children's Memorial Medical Center in Chicago. Step 2 was to identify and outline the anatomical structures of interest on each of approximately 200 images. These structures included the pelvis, sacrum,

Table 1
Characteristics of the cerebral palsy subjects who were imaged

	Subject 1	Subject 2	Subject 3
Gender	F	M	M
Age (years)	7	14	27
Diagnosis	Diplegia	Hemiplegia	Diplegia
Hip rotation, range of motion (°) (internal/external)	65/25	90/10	80/30
Hip flexion contracture (°)[a]	20	10	15
Popliteal angle (°)[b]	55	85	40
Anteversion angle (°)[c]	47	44	46

[a] Quantifies the degree to which the hip can be extended without arching the spine. This test is performed with the subject supine, the pelvis stabilized, and the contralateral limb flexed. Hip flexion contracture is determined by measuring the angle between the subject's extended limb and the examining table [30].

[b] Quantifies the degree to which the knee can be passively extended with the hip flexed 90°. This test is performed with the subject supine, the pelvis stabilized, and the contralateral limb flexed. Popliteal angle is defined as the angle between the shank and a vertical line. A popliteal angle greater than 45° may indicate hamstrings tightness [31].

[c] Defined as the angle between the plane of the femoral neck axis and the plane of the condylar axis [32]. This angle is 10–15° in unimpaired adults [33].

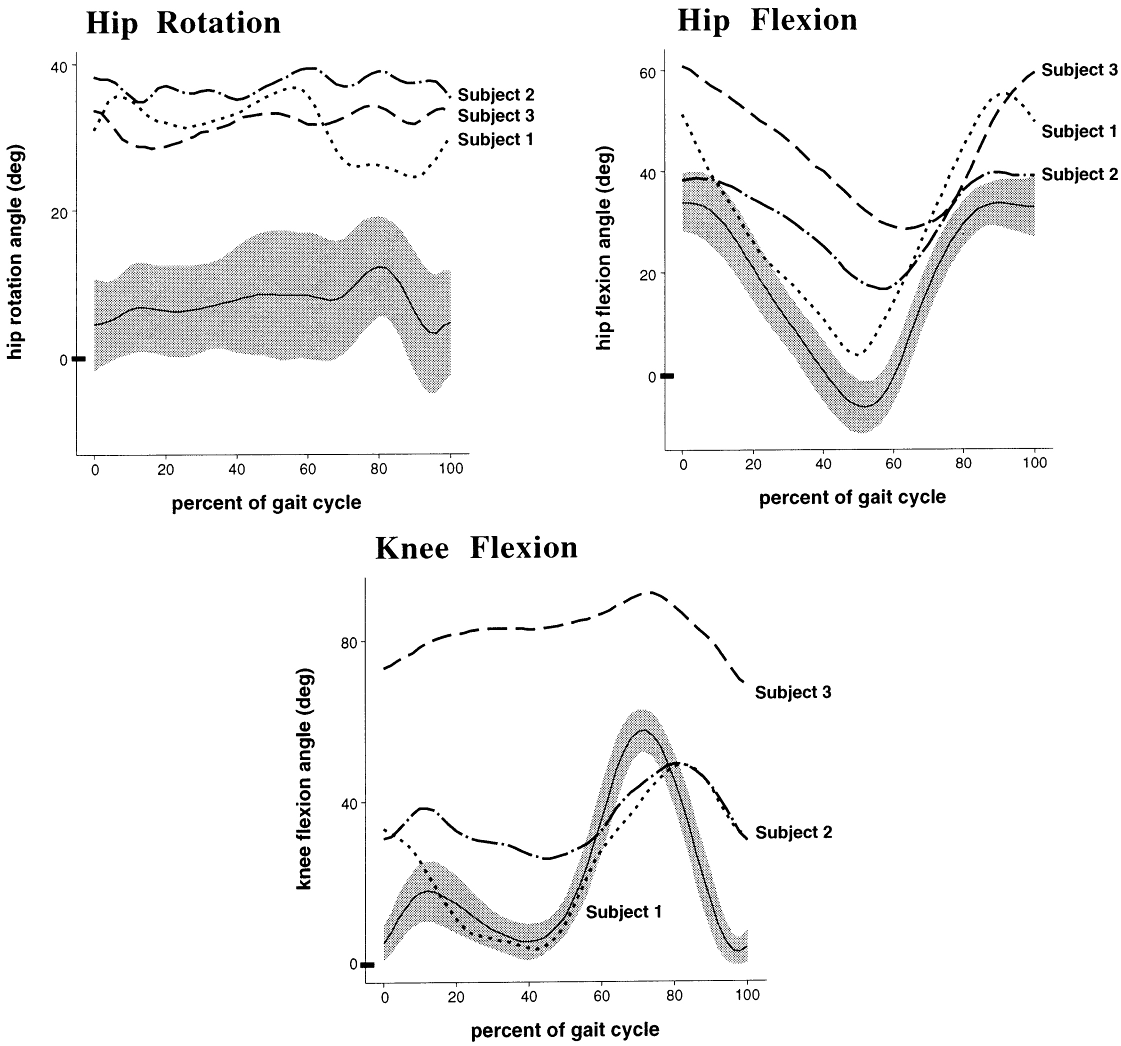


Fig. 1. Hip rotation, hip flexion, and knee flexion angles of the cerebral palsy subjects during walking. For each subject, data are displayed for the limb that was imaged (dashed lines). Averaged gait kinematics for 18 unimpaired subjects (mean ± 1 S.D., shaded areas) are shown for comparison.

femur, tibia, medial hamstrings and adductor muscles. Step 3 was to reconstruct the three-dimensional surfaces 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, three-dimensional representation of each subject's anatomy at one limb position, the position at which the subject was scanned. Step 5 was to define kinematic models of the hip and the knee based on the bone surface geometry. Step 6 was to characterize the paths of the muscles for a functional range of joint motions. The resulting models were capable of estimating muscle lengths and moment arms for a wide range of limb positions, including hip and knee angles that corresponded to gait.

2.1. Protocol for imaging

Each model was constructed from three overlapping series of transverse images and one or two series of sagittal images. In the transverse plane, images were taken in contiguous 3-mm intervals from the ilium to below the lesser trochanter, and from the superior pole of the patella to below the insertions of the hamstrings. Less accuracy was required along the femoral shaft; hence, scans were acquired at 10-mm intervals from the base of the femoral neck to the patella. In the sagittal plane, images of the knee were taken at 3-mm intervals. The imaging session lasted less than 2 h and was tolerated well by the subjects.

Imaging parameters were chosen to provide good spatial resolution and contrast, suitable for the development of detailed musculoskeletal models. The three series of transverse images were acquired using a body coil. The sagittal images were obtained using an extremity coil. To enable differentiation of the muscle boundaries in the images, a T1-weighted spin-echo pulse sequence was applied (TR, 400 ms; TE, 17 ms;

FOV, 20–24 cm; matrix, 256 × 256), which enhanced the signal from the fatty tissues and fascia surrounding each muscle.

During the imaging session, the subject was positioned supine with the hip and knee slightly flexed. Firm cushions were used to support the subject's limb, and a generous amount of tape was applied to help minimize movement. For the series of sagittal images, we attempted to secure the subject's limb at different knee flexion angles, corresponding to the maximum flexion that could be achieved in the extremity coil (25–35°), and to the maximum extension that could be attained by the subject. Two neoprene cuffs (TRS, Boulder, CO), each with a set of five vitamin E capsules attached, were fastened around the subject's thigh to facilitate registration of the overlapping series of images.

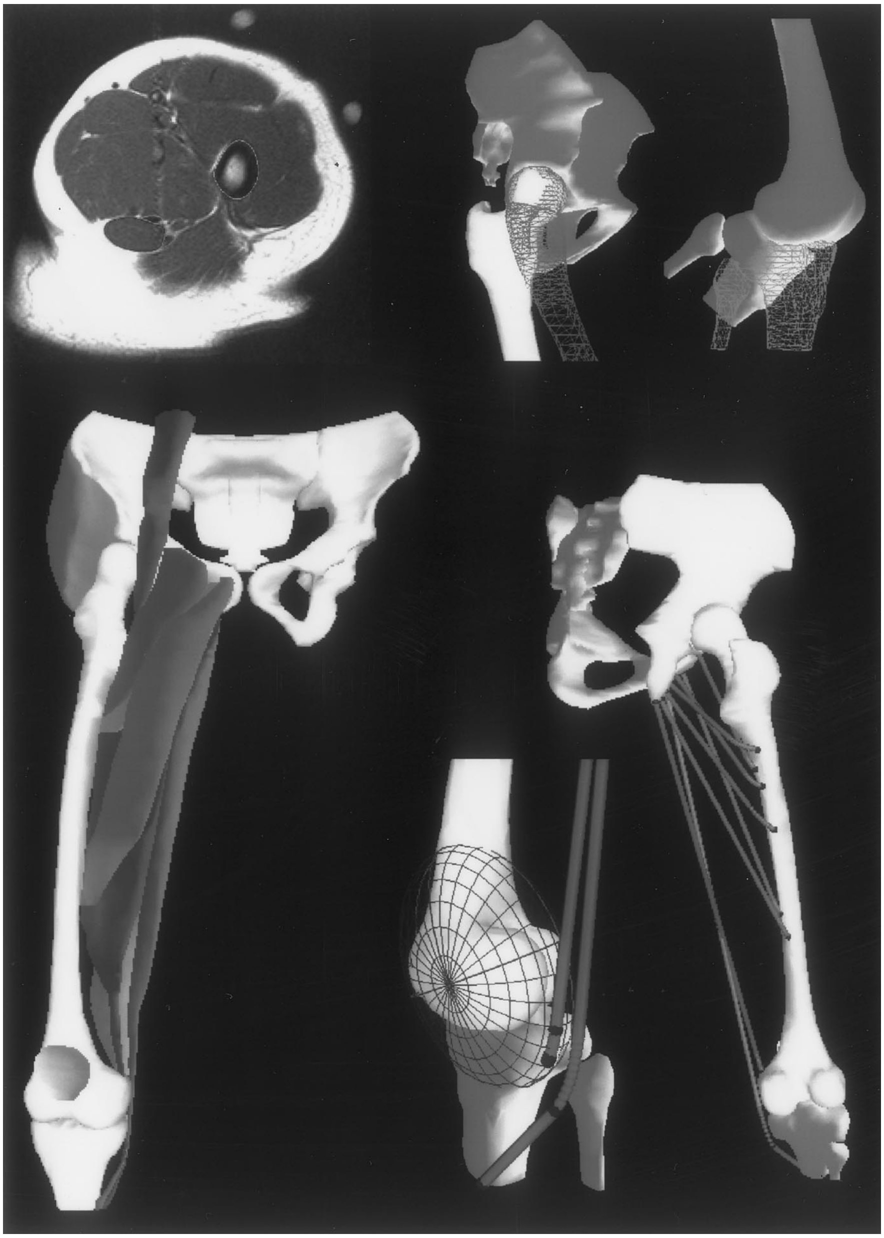


Fig. 2. Construction of a subject-specific model from magnetic resonance images. Three-dimensional surface models of the muscles and bones were generated from two-dimensional contours segmented manually from the images (top left). Surfaces from overlapping series of images were registered to obtain an accurate representation of the subject's anatomy at the 'scanned' limb position (bottom left). Kinematic models of the hip and the knee were implemented (top right), and the muscle lines of action were defined (bottom right). Ellipsoidal 'wrapping surfaces' were used to simulate wrapping of the medial hamstrings and gracilis over the gastrocnemius and posterior femoral condyles with knee extension. The adductors were prescribed to wrap around the femoral shaft and underlying muscles with hip internal rotation.

2.2. Methods for three-dimensional reconstruction and registration

Three-dimensional representations of the bones, muscles, and vitamin E capsules were generated by outlining the boundaries of each structure in each image slice. A three-dimensional surface model of each structure was created by connecting adjacent contours with a polygonal mesh.

Surfaces generated from overlapping series of images were registered in two steps. First, approximate transformations between adjacent series of images were obtained from the vitamin E capsules. Reference frames for each series were defined from the centroids of the capsules, and the coordinate transformations between sets of series were computed using a least-squares algorithm [12]. In the second step, the surface geometry of the muscles and bones was carefully inspected and used to refine the initial transformations.

2.3. Specification of joint kinematics

The surface meshes representing the muscles and bones were imported into a musculoskeletal modeling package, SIMM [13]. Coordinate systems for the pelvis, femur, and tibia were defined from anatomical landmarks, and kinematic descriptions of the hip and tibiofemoral joints were specified for each subject based on the bone surface geometry.

The hip was assumed to be a ball-and-socket joint. The hip center was estimated by fitting a sphere to the surface of the femoral head using a Gauss-Newton nonlinear least-squares algorithm (MATLAB, MathWorks, Natick, MA).

The knee joint specified the three-dimensional 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 [14,15]. Walker et al. reported tibiofemoral translations that had been scaled to a 'nominal-sized' adult knee and averaged for 23 specimens; however, the knees of our subjects with cerebral palsy varied considerably in size. For this reason, an iterative procedure was developed to estimate, for each subject, an appropriate scale factor which, when multiplied by the tibiofemoral translations specified by Walker et al., produced tibiofemoral contact locations that were consistent with experimental data published in the literature [16]. Since the relative positions of the femur and tibia were known from the image data at one or more angles of knee flexion, the scaled knee kinematics at every iteration were computed and evaluated based on these reference positions. The resulting tibiofemoral joint for each model prescribed the three-dimensional motions of the tibia relative to the femur for a 120° range of knee flexion.

2.4. Representation of muscle-tendon paths

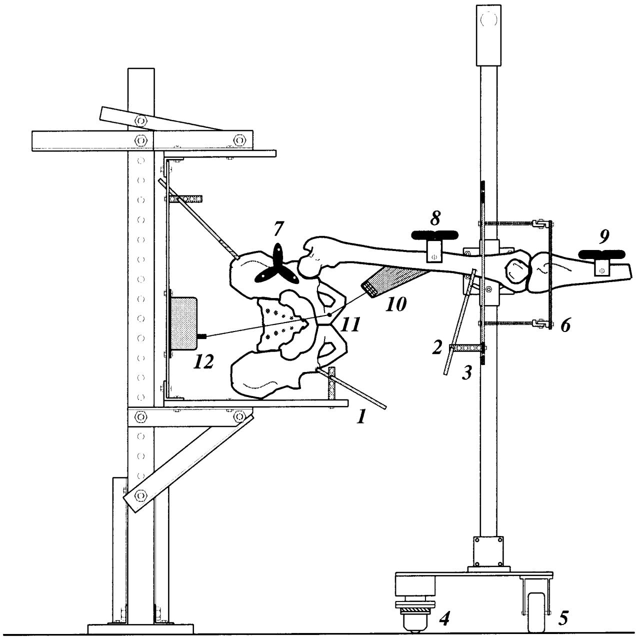
Each muscle was represented as a series of line segments from origin to insertion. The attachment sites were specified based on the muscles' three-dimensional surface representations at the scanned limb position. To define the muscle paths at other limb positions, ellipsoidal wrapping surfaces and via points [17] were introduced to simulate underlying structures and other anatomical constraints. The medial hamstrings and gracilis were defined to wrap over the gastrocnemius and posterior femoral condyles with knee extension, and the other adductors were specified to wrap around the femoral shaft and underlying muscles with hip internal rotation.

2.5. Evaluation of models based on tendon excursion measurements

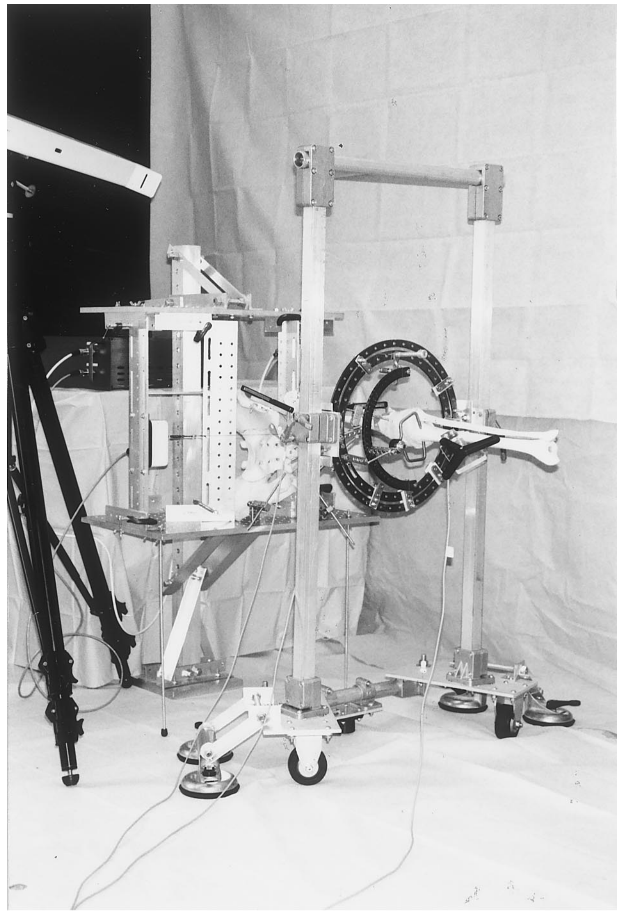
We performed a set of anatomical experiments to evaluate our methods for building the subject-specific models and, in particular, to quantify the accuracy of hip rotation moment arms estimated from these models. MR images of three lower extremity cadaveric specimens were acquired using imaging protocols and parameters similar to those used for the cerebral palsy subjects. From these images, a musculoskeletal model of each specimen was constructed. Muscle moment arms estimated from these models were compared to the moment arms determined on the same specimens from tendon excursion measurements. Specimen 1 was a female hemipelvis specimen, and Specimens 2 and 3 were complete male pelves. The specimens varied in size, with distances from hip center to knee center ranging from 38 cm (Specimen 1) to 43 cm (Specimen 3).

Moment arms of the three specimens were determined experimentally using the tendon excursion method [18]. In these experiments, changes in the origin-to-insertion lengths of the muscles were measured with internal and external rotation of the hip. Each specimen was prepared by removing the skin, subcutaneous tissues, and fascia. The muscles of interest were identified and cleaned, and other soft tissues were removed as needed. Tissues in the popliteal region which were thought to potentially influence the paths of the hamstrings or gracilis were preserved. The hip capsule and knee ligaments were left intact.

Screws were drilled into bony landmarks on the pelvis and femur to enable the repeatable definition of coordinate systems aligned with anatomical axes. A three-dimensional localizer (FlashPoint 5000, Image Guided Technologies, Boulder, CO) was used to digitize these landmarks and to track the positions and orientations of infrared emitter triads which were rigidly attached to the bones.



(a)



(b)

Fig. 3. Schematic (a) and photograph (b) of the experimental set-up to measure muscle-tendon excursions with hip rotation. Ilizarov components and half-pins (1, 2) were used to secure the pelvis in a jig and to mount the femur in a set of concentric rings (3) which ride on a cart. The rings could be raised or lowered on their support posts to adjust hip adduction. The inner ring could be fixed or rotated relative to the outer ring to control hip rotation. To vary hip flexion, the cart was equipped with one ball caster (4) and two rigid casters (5), which were mounted on adjustable plates. The plates were angled, according to femur length, so that the cart rolled in a circular arc. Detachable suction cups enabled the cart to be secured at any angle of hip flexion. A locking hinge (6) attached to the tibia provided control of knee flexion. Joint angles were monitored by tracking the locations of infrared emitter triads (7, 8, 9) mounted to the pelvis, femur, and tibia. Muscle-tendon excursions were measured by connecting a suture to a nylon mesh sewn at the distal tendon of each muscle (10), routing the suture through a suture anchor at the muscle's origin (11), and attaching it to a position transducer (12).

The specimen was mounted in a custom designed jig (Fig. 3) that provided control of hip rotation, hip flexion, and knee flexion. Four or five cortical bone screws (EBI Medical Systems, Parsippany, NJ) were used to fix the pelvis to the base plate of the jig. Additional half-pins and Ilizarov components [19] were used to secure the femur to the inner of two concentric rings separated by sleeve bearings, which were mounted on a cart. The inner ring could be clamped or rotated relative to the outer ring to vary the angle of hip rotation. The cart was equipped with one precision ball caster and two rigid casters, which were mounted on adjustable plates. The plates of the cart could be angled, depending on the distance between the specimen's hip center and the rings, so that the cart rolled in a circular arc about the specimen's hip center, thereby flexing the hip. Suction cups were used to secure the cart at desired angles of hip flexion. A locking hinge attached to the tibia provided control of knee flexion.

Accurate calculation of the muscle moment arms required careful alignment of the specimen in the jig. In particular, (1) the pelvis had to be fixed such that its medial–lateral axis was perpendicular to the base plate of the jig, (2) the femur had to be secured such that its superior–inferior axis passed through the center of the concentric rings, perpendicular to the plane of the rings, and (3) the adjustable plates of the cart had to be angled so that the rotation axis of the cart was collinear with the specimen's hip center. To achieve accurate alignment, custom software was written (LabVIEW, National Instruments Corporation, Austin, TX) and used in combination with the three-dimensional localizer to guide each of these steps. Hip flexion, adduction, rotation, and knee flexion angles were monitored during the experiment by tracking the locations of emitter triads mounted to the pelvis, femur, and tibia.

Tendon excursion measurements were made by sewing polyester suture to the distal tendon of each muscle, routing the suture through a suture anchor at the muscle's origin, and attaching it to a position transducer (PT101 precision potentiometer transducer, Celesco Transducer Products, Canoga Park, CA). The transducer applied a constant tension of 7.5 N, and was reported to be accurate to within 0.15% of full scale, or ± 0.38 mm. The data were sampled at 15 Hz using a 16-bit analog-to-digital converter (PCI-MIO-16XE-50, National Instruments Corporation, Austin, TX). The A/D board determined the resolution of the measurements, which was 0.04 mm.

The length changes of the medial hamstrings and adductors were measured as the femur was slowly rotated through a range of internal and external rotation. The muscle length changes that occurred with hip rotation were measured at hip flexion angles of 0, 25, 45, 60 and 90° and at knee flexion angles of 0, 30 and 60°. A minimum of five trials were collected for each test condition.

Moment arm curves were obtained from the derivatives of the tendon excursion versus rotation angle data. The numerical derivative of each trial was calculated, and a second-order Butterworth filter with a cut-off frequency between 1 and 2 radians^{-1} was applied (MATLAB, MathWorks, Natick, MA). The measured moment arms were averaged over multiple trials and were compared to the moment arms computed with the corresponding specimen-specific model.

3. Results

Rotational moment arms of the medial hamstrings, adductor brevis, adductor longus, and gracilis calculated from the specimen-specific models compared favorably to the moment arms determined experimentally on the same specimens (Table 2). For each of the specimens, errors in the moment arms computed with the model were less than 4 mm at the upright standing position, averaged over the range of hip rotation angles measured during the experiment. Similar errors were obtained for other limb positions. Based on these data, we are confident that our methods for constructing musculoskeletal models from MR images are suitably accurate for determining hip rotation moment arms in persons with cerebral palsy.

Examination of hip rotation moment arms in three subjects with cerebral palsy who walked with crouched, internally-rotated gait revealed that the semimembranosus, semitendinosus, and gracilis had negligible or external rotation moment arms throughout the gait cycle (Fig. 4). The adductor brevis and adductor longus had internal rotation moment arms in Subject 2, and the adductor longus had an internal rotation moment arm in Subject 3, but the moment arms of these muscles were small (< 4 mm). These results suggest that neither the medial hamstrings nor the adductor brevis, adductor longus, or gracilis are likely to contribute substantially to excessive internal rotation of the hip.

Table 2
Average errors[a] in hip rotation moment arms computed from specimen-specific models[b]

	Specimen 1	Specimen 2	Specimen 3
Semimembranosus	1.2	0.9	2.1
Semitendinosus	3.4	3.8	3.2
Adductor brevis	0.3	1.5	3.4
Adductor longus	0.3	1.6	2.8
Gracilis	3.8	NM[c]	1.6

[a] Error defined as the absolute value of the difference between the experimentally-determined moment arm and the moment arm computed with the model, averaged over the range of hip internal and external rotation angles achieved during the experiment. All errors are reported in units of mm.

[b] Moment arms computed with hip flexion, adduction, and knee flexion angles = 0°.

[c] NM, not modeled.

4. Discussion

Previous descriptions of hip rotation moment arms, based on anatomical experiments and computer models representing normal adult musculoskeletal geometry, have indicated that the medial hamstrings, adductor brevis, and adductor longus have the capacity to internally rotate the hip in the upright, standing position [6,8]. Some interventions commonly performed in persons with cerebral palsy, such as lengthening of the medial hamstrings and adductors, are expected to decrease the internal rotation moments produced by these muscles. However, the rotational functions of the hamstrings and adductors at body positions commonly assumed by patients during walking are unknown, and the effects of femoral deformities on the rotational moment arms of the muscles have not been determined. As a result, surgical lengthenings of the medial hamstrings and adductors intended to decrease internal rotation are currently planned without quantitative data that describe the rotational function of the muscles in persons with cerebral palsy.

In this study, we (1) developed models of three subjects with cerebral palsy from MR images, (2) tested the accuracy of the methods used to construct these models, and (3) used the models to determine the rotational moment arms of the medial hamstrings and adductors at limb positions corresponding to crouched, internally-rotated gait. We discovered that the rota-

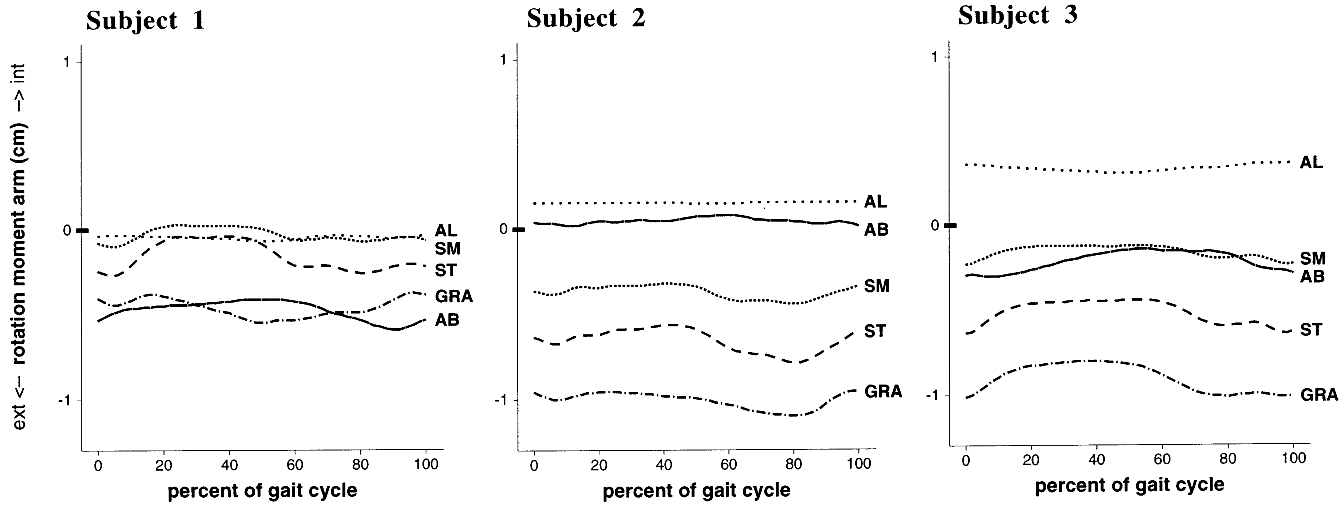


Fig. 4. Hip rotation moment arms of the adductor brevis (AB), adductor longus (AL), gracilis (GRA), semimembranosus (SM), and semitendinosus (ST) versus gait cycle for Subjects 1, 2, and 3. Moment arms were estimated using the subjects' measured three-dimensional gait kinematics in conjunction with the model of each subject constructed from MR images.

tional moment arms of the medial hamstrings, adductor brevis, adductor longus, and gracilis were either very small or external throughout the gait cycle in all three subjects. Hence, none of these muscles could have generated a substantial hip internal rotation moment in these subjects, whose gait abnormalities and femoral anteversion angles might be considered typical of patients who walk with excessive internal rotation of the hip.

To put these data into perspective, we compared the rotational moment arms of the medial hamstrings and adductors to the rotational moment arms of the anterior compartment of the gluteus medius during crouched, internally-rotated gait. A recent experimental study has shown that the internal rotation moment arms of the gluteus medius and gluteus minimus increase dramatically with flexion of the hip [20]. The moment arms of the gluteus medius calculated with our models of the cerebral palsy subjects are consistent with this observation (Fig. 5). In the model of Subject 3, for example, the rotational moment arm of the anterior compartment of gluteus medius is over four times the moment arm of the adductor longus. Since excessive flexion of the hip frequently accompanies internally-rotated gait [9,21], and since the gluteal muscles are typically active and play an important role in walking [22,23], we hypothesize that the excessive hip flexion of these patients, which increases the internal rotation moment arms of the gluteus medius and minimus, is more likely to cause internal rotation than the hamstrings or adductors.

It is important to keep in mind the limitations of this study. First, errors in the hip rotation moment arms of the hamstrings and adductors computed from models of three cadavers and averaged over a range of hip rotation angles, in some cases, were as much as 4 mm. The MR images acquired in vivo for the subjects with cerebral palsy were generally of better quality than the images obtained for the specimens; nevertheless, the rotational moment arms determined using the subject-specific models may be inaccurate by a few millimeters. These potential errors are the same order of magnitude as the moment arms of the muscles. However, because both the moment arms and the potential errors are small, we are confident that our clinical conclusions are not sensitive to such inaccuracies.

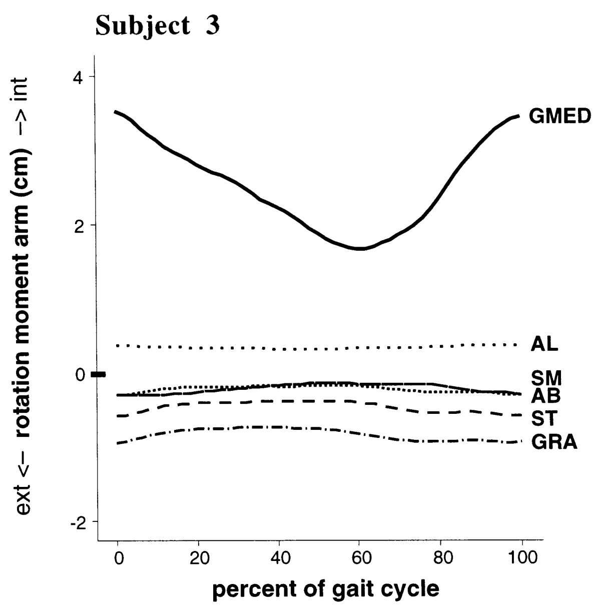


Fig. 5. Hip rotation moment arms of the medial hamstrings, adductors, and anterior compartment of the gluteus medius (GMED) during walking, calculated using the model of Subject 3 constructed from MR images.

Second, we have inferred the rotational functions of the medial hamstrings and adductors from static analyses of the musculoskeletal geometry. Our subject-specific models do 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 moments through dynamic coupling [24].

Third, patients with internal rotation gait frequently undergo derotational osteotomies of the femur to improve alignment of the limb. Alterations in femoral geometry that occur with this surgery may change the rotational moment arms of the muscles. While changes in the rotational function of muscles after surgery for the three subjects studied here are not reported, our previous studies suggest that the changes in the rotational moment arms of the hamstrings and adductors after derotational osteotomy are minimal [25].

Fourth, all three of the cerebral palsy subjects who were imaged walked with 30–40° of hip internal rotation throughout the gait cycle and had femoral anteversion angles of about 45°. Some patients who walk with a crouched, internally-rotated gait exhibit less hip internal rotation [26,27]; other patients have been reported to have anteversion angles greater than 60° [28,29]. Additional work is needed to understand how the rotational moment arms of the hip muscles vary with hip rotation, hip flexion, knee flexion, and femoral anteversion before our conclusions and clinical recommendations can be generalized to patients who exhibit much different gait patterns or a wider range of anteversion deformities than the subjects included in this study.

Despite these limitations, this study provides important evidence which suggests that the medial hamstrings and adductors do not contribute to excessive internal rotation of the hip in many persons with cerebral palsy who walk with crouched, internally-rotated gait. Therefore, surgical lengthening of the hamstrings or adductors in an attempt to reduce excessive internal rotation of the hip may be inappropriate. This analysis demonstrates the need for biomechanical guidelines, based on accurate descriptions of joint kinematics and musculoskeletal geometry, that can aid treatment planning for persons with cerebral palsy.

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