Pergamon

J. Biomechanics, Vol. 27, No. 10, pp. 1201–1211, 1994

0021–9290/94 $7.00+.00

0021–9290(94)E0012-R

TRANSFER OF THE RECTUS FEMORIS: EFFECTS OF TRANSFER SITE ON MOMENT ARMS ABOUT THE KNEE AND HIP

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Abstract—Decreased range of knee motion during gait is often treated by surgically releasing the rectus femoris from the patella and transferring it to one of four sites: semitendinosus, gracilis, sartorius, or the iliotibial tract. This study was conducted to determine if there are differences between these four tendon transfer sites in terms of post-surgical moment arms about the knee and hip. A graphics-based model of the lower extremity was used to simulate the origin-to-insertion path of the rectus femoris after transfer. Anatomical studies were conducted to evaluate the accuracy of the simulated tendon transfers by comparing knee flexion moment arms calculated with the computer model to moment arms measured in two anatomical specimens. The computer simulations and anatomical studies revealed substantial differences in the knee moment arms between the four sites. We found that the rectus femoris has the largest peak knee flexion moment arm (4–5 cm) after transfer to the semitendinosus. In contrast, after transfer to the iliotibial tract the rectus femoris has a slight (0–5 mm) knee extension moment arm. None of the transfers to muscle–tendon complexes on the medial side of the knee (semitendinosus, gracilis, sartorius) substantially affect the hip rotation moment arm of the rectus femoris. Transferring to the iliotibial tract increases hip internal rotation moment arm of the rectus femoris, but only when the hip is externally rotated.

INTRODUCTION

Patients with spastic cerebral palsy often undergo orthopaedic surgeries aimed at improving their gait abnormalities. One of the most common abnormalities, crouch gait, is characterized by excessive knee flexion during the stance phase. Excessive knee flexion during stance can usually be reduced by surgically elongating the tendons of the hamstrings (Gage *et al.*, 1987; Ray and Ehrlich, 1979; Thometz *et al.*, 1989). However, a common and serious complication of isolated hamstring lengthening is decreased postoperative knee flexion during the swing phase (Baumann *et al.*, 1980). Inadequate knee flexion during swing interferes with toe clearance and can result in conspicuous and inefficient compensatory movements, such as circumduction of the hip.

The reputed cause of decreased knee flexion in the swing phase is over-activity of the rectus femoris muscle (Gage *et al.*, 1987; Perry, 1987). The rectus femoris, which produces a flexion moment about the hip and an extension moment about the knee, is normally active for a short burst at the beginning of the swing phase. In some patients with stroke or cerebral palsy, however, the rectus femoris has prolonged or continuous activity during the swing phase, which restricts knee flexion (Perry, 1987; Sutherland *et al.*, 1990; Waters *et al.* 1979).

Perry (1987) suggested transferring the rectus femoris in patients with prolonged swing phase activity to treat decreased swing phase knee flexion after hamstring lengthening. The surgery is now done in conjunction with hamstring lengthenings to avoid decreased knee flexion postoperatively. This tendon transfer is accomplished by surgically releasing the distal attachment of the rectus femoris from the patella, freeing it from surrounding tissue, and reattaching it to one of several sites posterior to the knee.

Sutherland *et al.* (1990) reported that swing phase knee flexion increased an average of 16° after transfer of the rectus femoris in patients with decreased swing phase knee flexion. Gage *et al.* (1987) reported that postoperative knee range of motion was greater during gait (i.e. stance phase knee flexion was decreased and swing phase knee flexion was increased or maintained) when hamstring lengthenings were performed in combination with the rectus femoris transfer. Ounpuu *et al.* (1993a) reported that swing phase knee flexion was maintained when hamstring lengthenings were performed in combination with the rectus femoris transfer. Because of this documented success, the rectus femoris transfer has become widely used to avoid or correct inadequate knee flexion during the swing phase of gait.

In addition to affecting knee motion, Gage (1990) suggested that the rectus femoris transfer may also

Received in final form 9 January 1994.
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help correct rotational deformities at the hip. He suggested that transfer to one of the medial knee flexors (e.g. semitendinosus) may create an external rotation moment about the hip that could help correct excessive internal rotation of the hip, which occurs in many patients with cerebral palsy. Gage (1990) also suggested that lateral transfer, to the iliotibial tract, would produce an internal rotation moment to help correct excessive external rotation of the hip.

The distal tendon of the rectus femoris is typically transferred to one of four sites: the semitendinosus, the gracilis, the sartorius, or the iliotibial tract. The objective of this study was to determine if there are differences between these four commonly used transfer sites in terms of the postoperative moment arms at the knee and hip. We believe that accurate descriptions of the postoperative geometry using computer models that are tested with quantitative anatomical experiments will provide a basis for more effective tendon transfer design.

METHODS

A computer model of the human lower extremity (Delp *et al.*, 1990) was used to study the rectus femoris transfer. This model specifies the origin-to-insertion path of each muscle-tendon complex as a series of line segments so that muscle moment arms and lengths can be computed over a range of body positions. Using a computer graphics workstation, the path of any muscle–tendon complex can be altered by graphically adjusting its origin, insertion, or intermediate 'via points' to stimulate the geometric changes that occur during a tendon transfer.

To ascertain how the musculoskeletal geometry of the rectus femoris transfer could be accurately represented, an anatomical study was conducted during which the origin-to-insertion path of the rectus femoris was digitized after the muscle was transferred to each of the four post-surgical sites. The left leg of one alcohol preserved cadaver was prepared for digitizing by making an incision from the anterior superior iliac spine to the patella. An incision this large would not be made in an actual surgery; however, it was necessary to expose the entire muscle–tendon complex so that it could be digitized. The skin was removed, but the fascia surrounding the muscles was left intact on the upper two thirds of the thigh to maintain normal anatomical constraints. As in an actual surgery, the distal third of the rectus femoris was freed from surrounding tissue and transferred, as described by Gage *et al.* (1987). Once the distal end of rectus femoris was attached to one of the tendon transfer sites, tacks were inserted along the muscle between the origin and the new insertion to mark the points that would be digitized. Indentations were made in the head of each tack so that the tip of the digitizer could be placed at the same point during each digitizing trial. The knee and hip were maintained in 0° flexion.

An OPTOTRAK/3010 digitizing system (Northern Digital, Waterloo, Canada) was used to collect the coordinates that describe the path of rectus femoris after transfer. Three cameras mounted on a rigid frame tracked the three-dimensional position of infrared emitting diods (IREDs). A digitizing probe that houses sixteen IREDs and software that extrapolates the three-dimensional coordinates of the probe's tip were used to record the position of the center of the tacks along each muscle–tendon path. The anatomy table that held the cadaver was locked in place approximately three meters from the cameras. At this distance, calibration tests showed that the position of the tip of the digitizing probe could be determined within 1 mm. A fixture with four IREDs was mounted to the table and scanned by the cameras to define a reference frame for the table. IREDs were mounted on the cadaver on the anterior superior iliac spines and tibial tuberosity to establish a reference frame for the lower extremity and to verify that the specimen did not move during data collection.

Digitized coordinates that describe the rectus femoris path after transfer to each site were transformed and displayed on the graphics workstation (Fig. 1). The coordinates were first transformed into the pelvic reference frame of the lower-extremity model. The origin of the transferred rectus femoris was then translated to the origin of the rectus femoris of the model. Since the dimensions of the cadaver were slightly different from the dimensions of the model, the coordinates describing the digitized muscle–tendon paths were scaled using a procedure similar to the method suggested by Brand *et al.* (1982). For each axis (anterior–posterior, medial–lateral, superior–inferior) a scale factor was computed as the ratio of a skeletal dimension of the model and the anatomical specimen (Table 1 lists the skeletal dimensions and scale factors). After scaling, each point in the digitized muscle–tendon path was attached to the appropriate body segmental reference frame. The origin was attached to the pelvis, the insertion to the tibia, and intermediate points to the femur.

Since the model characterizes the kinematics of the hip and knee, moment arms of rectus femoris after transfer to each site can be computed for a range of body positions. The hip is represented as a ball-and-socket joint; thus, moment arms can be computed for flexion, abduction, and rotation. The knee model accounts for the kinematics of the tibiofemoral joint and the patellofemoral joint in the sagittal plane; thus, only flexion–extension moment arms are computed (Delp *et al.*, 1990). The moment arm (MA) of the rectus femoris after transfer was computed as the partial derivative of muscle–tendon length (∂l) with respect to a joint angle ($\partial\theta$). That is, $\mathrm{MA} = \partial l/\partial\theta$ (An *et al.*, 1984). Knee flexion moment arm vs knee angle curves were calculated for each of the four transfer sites. The effects of transferring rectus femoris on hip rotation, hip abduction, and hip flexion moment arms were also evaluated using the model.

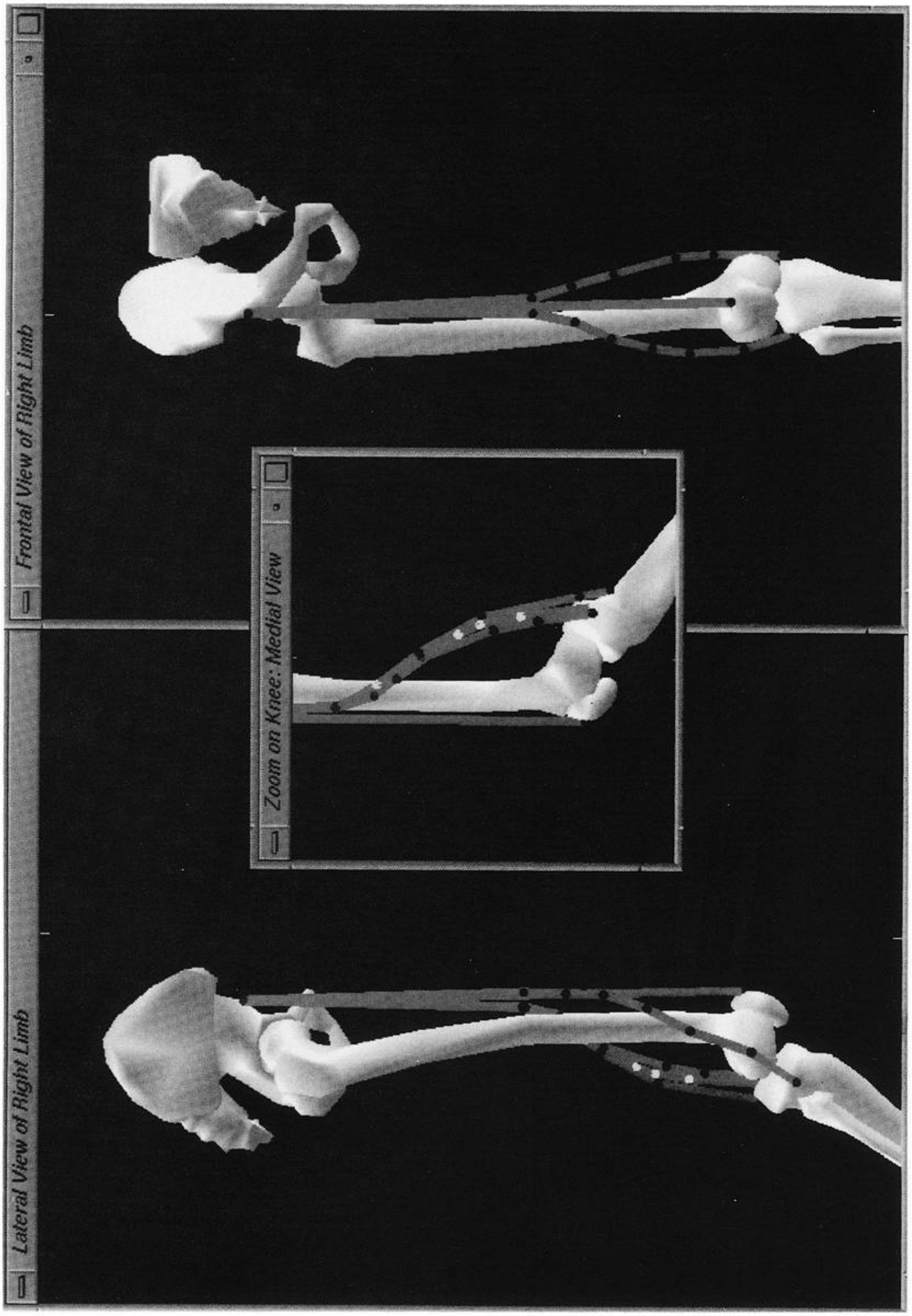

Fig. 1. Biomechanical model of the rectus femoris before and after transfer to four different sites. The left window shows a lateral view. The inset shows a medial view. The right window shows an anterior view.

Table 1. Skeletal dimensions used to determine scale factors

Skeletal dimension	Specimen	Model	Axis for scaling	Scale factor†
Anterior–posterior dimension of lateral femoral condyle	61	58	anterior–posterior	0.95
Femoral length: greater trochanter to lateral epicondyle	359	379	superior–interior	1.06
Maximum medial–lateral dimension of distal femur	76	80	medial–lateral	1.05

*All dimensions are in mm.
†Scale factor is computed as model dimension/specimen dimension.

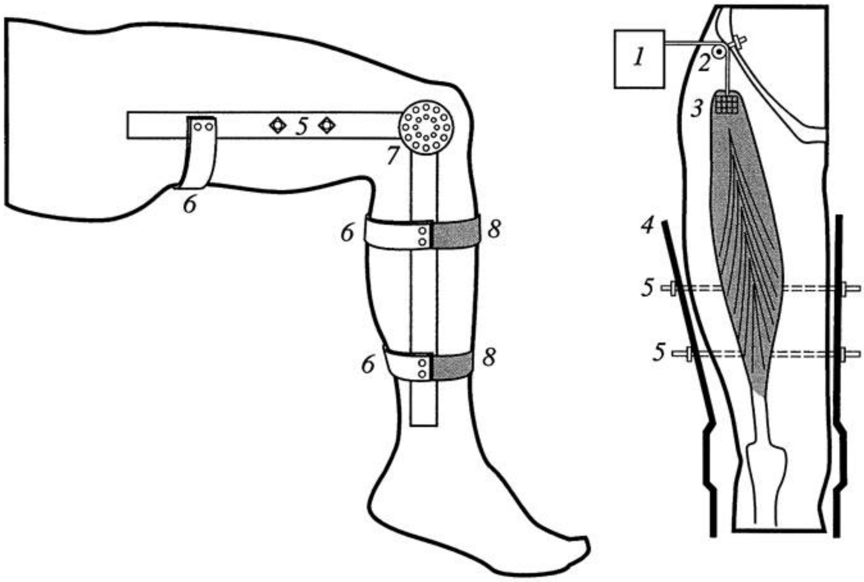

Fig. 2. Schematic diagram of the experimental setup used to measure tendon excursion as a function of knee flexion angle. The components of the system are: (1) position transducer, (2) pulley secured at the origin of the rectus femoris, (3) nylon mesh used to attach position transducer to the rectus femoris, (4) knee brace, (5) metal pins inserted through the femur, (6) metal cuffs riveted to brace, (7) locking hinge used to fix the knee and specific angles of flexion, (8) Velcro straps that help keep the leg and brace together, but allow sliding of the tibia relative to the brace.

To test the computer model, knee flexion moment arms were measured in two alcohol preserved cadavers by recording tendon displacement over a range of joint angles (An *et al.* 1984). One specimen was the right leg of a 90-year-old female and the other was the left leg of a 82-year-old male (see Table 2 for morphometric data that describe the specimens and model). Neither specimen had arthritic joints or severley atrophic muscles. Three incisions were made in the thigh to accommodate the four transfer sites: one along the anterior midline extending proximally 10 cm from the patella, one posteromedially extending proximally 10 cm from the adductor tubercle, and one laterally extending proximally 10 cm from the lateral epicondyle. The semitendinosus and gracilis transfer sites were prepared by cutting each tendon 8 cm proximal to its tibial insertion to provide sufficient length for the rectus femoris attachment. A piece of flexible nylon screen was sewn around sartorius, and the rectus femoris tendon was attached to the screen. The rectus femoris was attached to the iliotibial tract through a slit in its posterior border, approximately 6 cm, proximal to the lateral epiondyle. To measure tendon displacement it was necessary to free the rectus femoris from surrounding fascia along its entire length so that translation of the muscle could occur.

The rectus femoris was released from its origin and sewn to a piece of stiff nylon mesh (Fig. 2, note 3). A wire was connected to the mesh and fed through a pulley placed at the muscle's origin. This pulley insured that the muscle translated along its natural line of action. The free end of the wire was attached to a Celesco PT101 position transducer (Celesco Transducer Products, Canoga Park, CA), which is capable of ± 0.3 mm accuracy, ± 0.02 mm resolution, and loads the muscle with a constant tension of 7.5 N. The distal end of rectus femoris was attached to each tendon transfer site, and tendon displacement was measured over a range of knee flexion angles.

A locking brace mounted to the leg controlled the knee flexion angle during the tendon displacement measurements (Fig. 2). Two rods were drilled through

Table 2. Morphometric data for the two specimens and the model

	Female specimen	Male specimen	Model
Distance between anterior superior iliac spines	238	222	240
Distance between anterior inferior iliac spines (AIIS)	210	200	194
Distance between AIIS and lateral femoral epicondyle	419	483	462
Femoral length: greater trochanter to lateral epicondyle	359	410	379
Maximum anterior–posterior dimension of lateral femoral condyle	61	62	58
Maximum anterior–posterior dimension of medial femoral condyle	58	58	55
Maximum medial–lateral dimension of distal femur	76	85	80
Tibial length: tibial tuberosity to lateral malleolus	305	352	347
Maximum anterior–posterior dimension of lateral tibial condyle	43	48	45
Maximum anterior–posterior dimension of medial tibial condyle	47	50	49
Maximum medial–lateral dimension of tibial condyle	72	76	68

*All dimensions are in mm.

the femur to fix the upper segment of the brace. The lower segment of the brace was loosely strapped to the calf so that normal knee kinematics were not disrupted. A dial at the center of the brace's hinge indicated the knee flexion angle. The dial consisted of two rows of holes so that the brace could be locked at: 0° (full extension), 9, 15, 24, 30, 39, 45, 54, 60, 69, and 75° flexion by placing a pin through the corresponding hole in the dial. The lower leg was moved through the entire range of motion under constant load several times prior to the recording of tendon displacement for each transfer to facilitate the sliding of the tendons and to reduce the effects of creep.

Before the rectus femoris was transferred, tendon excursion was measured over a range of knee flexion angles with the distal tendon attached to its normal insertion on the patella. Following these normal trials, the distal tendon of the rectus femoris was attached to each transfer site and a total of 12 sets of tendon displacement vs knee angle data were recorded for each site. The knee angles were sequenced such that the movement lengthened the muscle–tendon complex because preliminary studies showed that this sequence decreased variability between trials. Thus, when the rectus femoris was attached to the patella (i.e. before transfer) the transducer reading with the knee in extension was assumed to be the zero displacement reading and was subtracted from all other readings in that trial. When the rectus femoris was attached to one of the knee flexors (i.e. semitendinosus, gracilis, sartorius), the excursion at maximum knee flexion (75°) was considered to be the zero displacement reading. The standard error was calculated to evaluate the variability between trials for each transfer site. The tendon displacement vs knee angle data for each site were fit with a fourth-order polynomial that minimized the sum of the squared residuals. The moment arm vs knee angle curve for each site was calculated by analytically differentiating the polynomial fit of the tendon displacement vs knee angle data.

RESULTS

The tendon excursion vs knee flexion angle curves are qualitatively similar for the male and female specimens (Fig. 3). The standard error calculated from the tendon displacement data was less than 0.2 mm for all experimental conditions except for one, the male normal condition, which had a maximum standard error of 0.4 mm. The square of the correlation coefficients (i.e. the R^2 values) are greater than 0.991 for all the polynomial fits, indicating that the fourth-order polynomials represent the tendon displacement data well.

There are substantial differences in the knee flexion moment arms between the four transfer sites (Fig. 4). Results from the model and the two specimens show that the knee flexion moment arm is greatest after transfer to the semitendinosus, followed by gracilis and sartorius. When rectus femoris is transferred to semitendinosus the peak knee flexion moment arm is approximately 4.5 cm, when transferred to gracilis the peak is approximately 3.5 cm, and when transferred to sartorius the peak is approximately 2.5 cm. The moment arm vs knee angle curves measured in the male and female specimens have the same shape for each transferred case, although the peak moment arm is consistently greater (by ≈ 5 mm) for the female specimen than the male specimen. The magnitudes of the peak knee flexion moment arms computed with the model are within 2 mm of the male specimen. However, the peak moment arm calculated with the model differs by as much as 8 mm from the peak moment arms measured in the female specimen. The knee flexion moment arm curves calculated with the model have the same relative magnitudes as the measured moment arms (i.e. semitendinosus has the largest knee flexion moment arm, followed by gracilis and sartorius). However, the moment arms calculated with the model peak at a more flexed knee position and vary less with knee angle than the measured moment arms.

The moment arm of rectus femoris after transfer to the iliotibial tract is very small for both specimens and the model (Fig. 4, dashed line). The moment arms measured in the two cadavers indicate that the rectus femoris has a slight (0–5 mm) knee extension moment arm when transferred to the iliotibial tract. The moment arm estimated with the model varies between a 6 mm extension moment arm with the knee extended

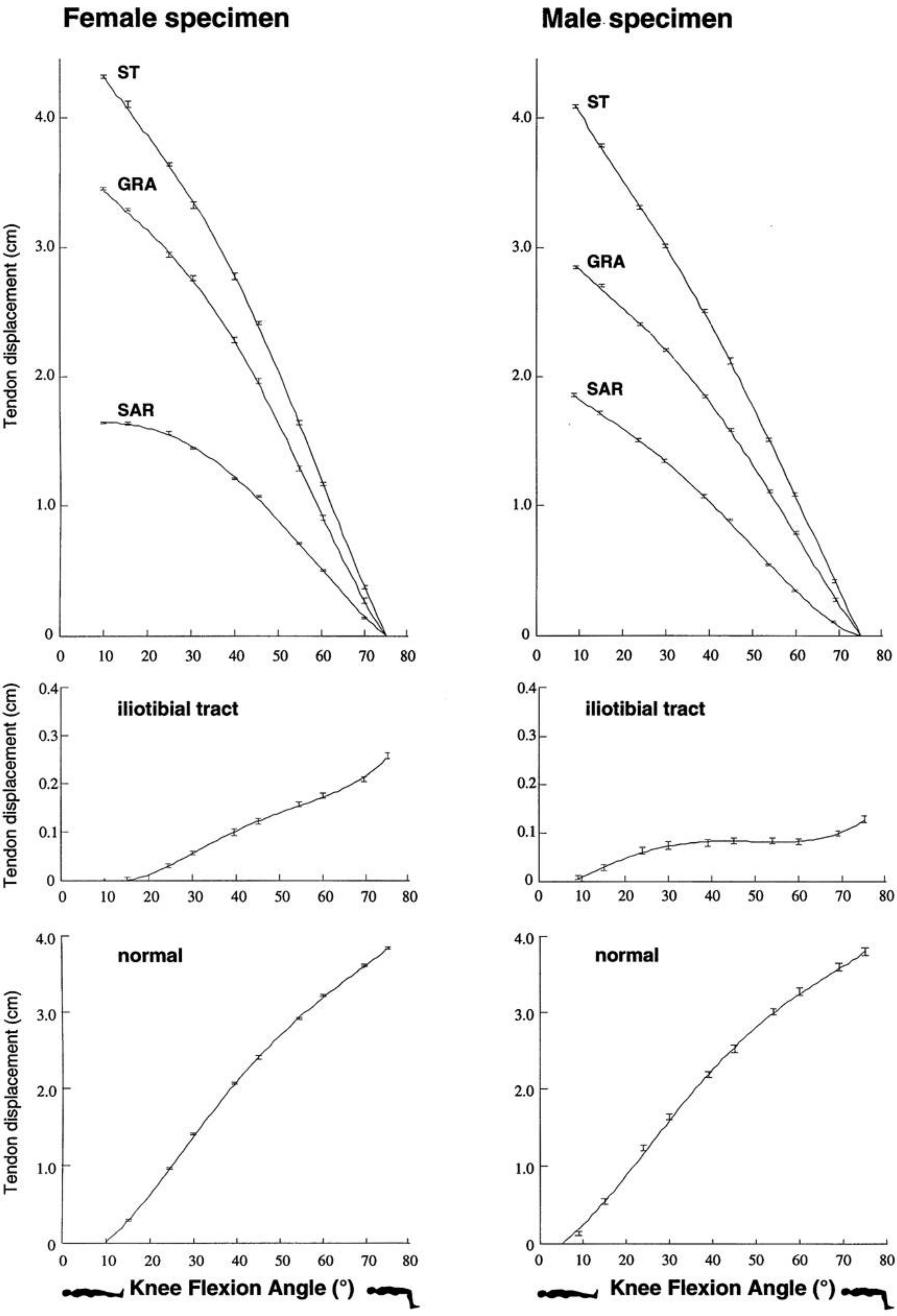

Fig. 3. Tendon displacement vs knee angle measured before transfer (normal) and after transfer to the semitendinosus (ST), gracilis (GRA), sartorius (SAR), and iliotibial tract. The curves show the fourth-order polynomial fits to the tendon displacement vs knee angle angle data. The error bars indicate ± one standard error of the measured tendon displacement, and show that there was little variability between trails.

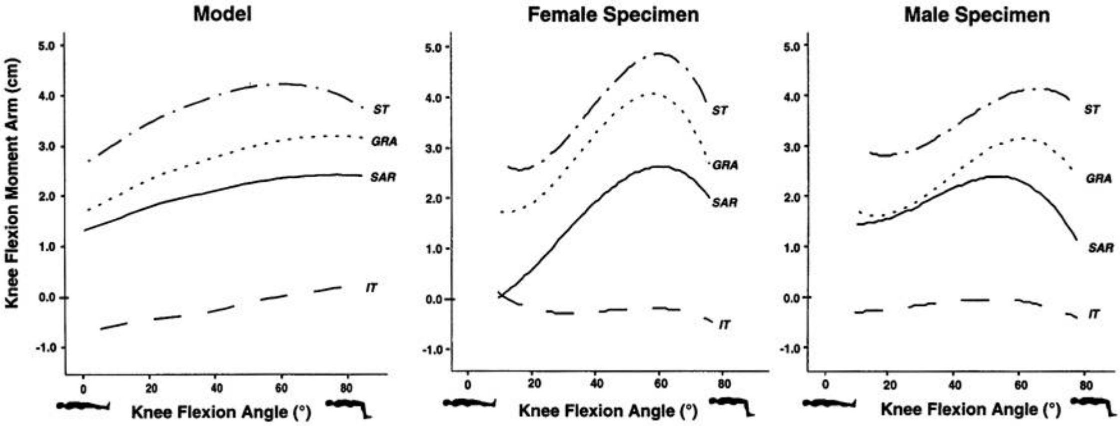

Fig. 4. Knee flexion moment arm vs knee flexion angle of the rectus femoris after transfer to each of the four sites. The curve marked ST is the moment arm after transfer to semitendinosus. GRA is the moment arm after transfer to gracilis. SAR is the moment arm after transfer to satorius. IT is the moment arm after transfer to the iliotibial tract. Positive moment arms indicate knee flexion. Negative moment arms indicate knee extension. Note that there are substantial differences in the moment arms between the transfer sites.

to a 2 mm flexion moment arm with the knee flexed.

Hip rotation moment arms of the rectus femoris are small (< 5 mm) both before and after transfer (Fig. 5). Since the differences between hip rotation moment arms after transfer to each of the medial knee flexors (semitendinosus gracilis, sartorius) were less than 1 mm, the medial transfers are presented as a group (thick shaded line). Hip rotation moment arms before transfer and after medial transfer are very similar (cf. solid and shaded lines in Fig. 5). Over a 30° range of hip rotation (10° external rotation to 20° internal rotation) the medial transfers and the normal rectus femoris have a small external rotation moment arm; the external rotation moment arm becomes larger (i.e. more negative in Fig. 5) with internal rotation. Transfer to iliotibial tract results in an internal rotation moment arm when the hip is externally rotated and an external rotation moment arm when the hip is internally rotated. Thus, this transfer tends to return the hip to the neutral position.

Transferring rectus femoris to any of the tendon sites has very small (< 2 mm) effect on hip flexion moment arm. Hip abduction moment arm decreases approximately 1 mm with medial transfer and increases about 1 mm with lateral transfer. The changes are small because the line of action of the proximal part of the muscle changed very little after transfer, since it is constrained by the fascia latae.

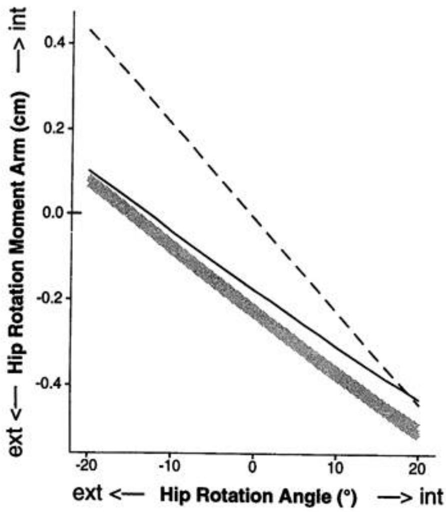

Fig. 5. Hip rotation moment arm vs hip rotation angle of the normal rectus femoris (solid line), after transfer to the iliotibial tract (dashed line) and after transfer to the medial knee flexors (thick shaded line). The rotational moment arms after transfer to the medial knee flexors are grouped since there is little difference between them (width of shaded line shows the range). The hip was flexed 30°. Positive angles and moment arms represent internal rotation; negative angles and moment arms represent external rotation. Note that the medial transfers have very little effect on the hip rotation moment arm.

DISCUSSION

The objective of this study was to determine if there are differences between four commonly used rectus femoris transfer sites in terms of moment arms at the knee and hip. Both computer simulations and anatomical studies showed that postoperative knee flexion moment arms depend strongly on transfer site. Our results indicate that the rectus femoris has the largest knee flexion moment arm after transfer to the semitendinosus. In contrast, rectus femoris has a slight

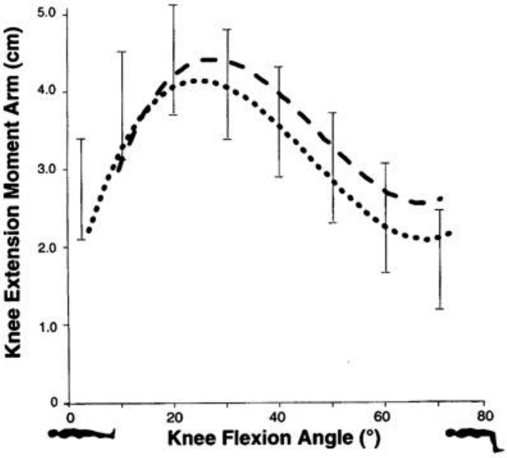

Fig. 6. Knee extension moment arm vs knee flexion angle measured before rectus femoris transfer. Moment arms measured in the male (dotted curve) and female (dashed curve) specimens correspond well with the effective knee extension moment arm reported by Grood *et al.* (1984). The error bars indicate the range of data reported by Grood *et al.* (1984, Fig. 8).

knee extension moment arm after transfer to the iliotibial tract. Moment arms about the hip were changed very little by transferring the rectus femoris.

Several modeling assumptions must be considered when using our results to understand knee flexion moment arms in patients who walk with decreased range of knee motion. First, the lower-limb model and anatomical specimens used in this study represent adult geometry. However, cerebral palsy patients who undergo tendon transfer surgeries are typically children. Although one would expect children to have smaller knee flexion moment arms than adults, due to smaller bone dimensions, the relationships between the knee flexion moment arms presented here are likely to be similar to moment arms in children. We make this statement based on the observation that our calculated and measured moment arms are consistent with anatomy. The sartorius inserts on the tibia more anteriorly (i.e. closer to the knee center of rotation) than gracilis or semitendinosus and should therefore have the smallest knee flexion moment arm of the three medial transfers. Gracilis inserts on the tibia between sartorius and semitendinosus and should therefore have a moment arm that is greater than sartorius, but less than semitendinosus. Semitendinosus has the most posterior insertion on the tibia and should therefore have the largest knee flexion moment arm, as our results indicate.

Second, our model and the anatomical specimens represent normal musculoskeletal geometry. However, patients with movement disabilities sometimes have bony abnormalities that could affect moment arms about the knee and hip. If the distal femur, the proximal tibia, and the relationships among the muscles about the knee are disrupted, the relative magnitudes of the knee flexion moment arms presented here may not be accurate. Also, deformities at the hip, such as subluxation of the femoral head, femoral anteversion, or a valgus femoral neck may affect the rotational moment arms of muscles crossing the hip. We therefore caution against using our results to understand the relationships between moment arms at the knee and hip in patients with substantial bony deformities.

Third, we have represented the transferred rectus femoris as an independent actuator, which transmits force only through its own tendon and not through surrounding musculature. Thus, if the distal tendon passes posterior to the instantaneous axis of rotation (e.g. after transfer to the semitendinosus) we assume that the rectus femoris generates a flexion moment about the knee when activated. However, fascial attachments between the proximal part of the rectus femoris and the surrounding vasti that remain following surgery may allow transmission of force through the remaining knee extensors. The extent to which muscles act as independent actuators remains an important question concerning the function of muscles after tendon transfer.

Fourth, even if one knew exactly how a tendon transfer affected muscle moment arms, the outcome of the surgery would still be unpredictable. To predict the outcome of a surgery, one would have to take into account many complicating factors, such as effects from other surgeries, changes in muscle activation patterns, and growth and adaptation of the musculoskeletal system. Such outcome predictions are beyond the scope of the current paper. This study focused on the moment arms of the rectus femoris because they are affected directly by the tendon transfer surgery, can be characterized well using computer modeling and can influence post-surgical joint motion.

Fifth, when muscle–tendon paths were digitized, only the distal part of the rectus femoris was moved because the proximal part of the muscle was held in place by the fascial sheath, as in an actual surgery (Fig. 1, right window). Hence, the moment arms of the rectus femoris about the hip changed very little after the transfer. Although the digitized muscle–tendon paths accurately represent musculoskeletal geometry immediately following surgery, the path of the transferred rectus femoris may change over time. Medial transfers to the semitendinosus, gracilis, and sartorius may cause medial remodeling of the proximal part of the rectus femoris, which would be bounded by sartorius. Similarly, transfer to the iliotibial tract may result in a lateral shift of the proximal part of the muscle, with the position of the iliotibial tract marking the lateral extreme.

To evaluate the effects of remodeling after tendon transfer, the digitized muscle–tendon paths of each transfer were modified on the computer graphics model. The paths of the rectus femoris after medial transfers were adjusted so that the proximal part of the paths were positioned just lateral to sartorius. This allowed analysis of the maximum possible changes that could be caused by medial remodeling. The effects of medial remodeling were small. The external rotation moment arm increased approximately 2 mm, and abduction moment arm of the rectus femoris decreased approximately 3 mm. The effects of lateral remodeling were evaluated by shifting the proximal part of the muscle–tendon path of rectus femoris after transfer to the iliotibial tract so that it was positioned just medial to the iliotibial tract. Lateral remodeling increased the range of hip rotation angles over which the rectus femoris acts as an internal rotator and increased the magnitude of the internal rotation moment arm approximately 5 mm, with the hip in external rotation. Lateral remodeling also increased the abduction moment arm approximately 4 mm.

Several sources of potential experimental error must also be considered. We estimated knee moment arms using the tendon displacement method, although several other methods are available. For example, Grood *et al.* (1984) calculated quadriceps moment arms by dividing a measured patellar tendon force into an applied knee moment. The knee extension moment arms we estimated from tendon excursions for the normal rectus femoris (i.e. with the rectus femoris attached to the patella) are similar to the moment arms reported by Grood *et al.* (1984) (Fig. 6). Thus, the moment arms estimated from measurements of tendon excursion appear to be reasonable.

Relative movement of the brace and the leg, which could create differences between the brace angle and the actual knee angle, is a second potential source of experimental error. However, given that the standard error in the tendon excursion curves are small (Fig. 3), we believe that these error sources do not adversely affect the quality of our tendon excursion data. The fit of the excursion data can also affect the moment arm estimates. Since the fits of the tendon excursion vs knee angle data are excellent, the derivatives of the fits are good approximations of the muscle moment arms.

Finally, errors in digitizing the muscle–tendon paths could lead to inaccurate moment arm calculations. Three trials of each tendon transfer path were recorded to quantify the differences in the digitized coordinates. Each digitized point in trial 1 was compared to the corresponding point in trial 2 by calculating the distance between the points. Likewise, trial 1 was compared with trial 3, and trial 2 with trial 3. The maximum distance between any two points are 3 mm, which occurred primarily in the anterior–posterior dimension for points in the upper thigh. The effect of this variation on calculated moment arms was minimal.

Even with the limitations discussed above, the results presented here have several important clinical implications. Children with cerebral palsy often exhibit rotational abnormalities at the hip in addition to decreased knee range of motion. The results presented here show that transferring rectus femoris medially to sartorius, gracilis, or semitendinosus has a minimal effect on the hip rotation moment arm of the rectus femoris. Transferring the rectus femoris to the iliotibial tract increases internal hip rotation moment and may be effective in correcting a hip that is externally rotated. This result is consistent with the finding of Gage *et al.* (1987) that patients with an external rotation deformity showed a decrease in external rotation when rectus femoris was transferred to the iliotibial tract.

There are several possible means by which the rectus femoris transfer can increase the range of knee motion during swing. The excessive knee extension moment generated by an over-active rectus femoris during the swing phase of gait may be eliminated by the transfer. If this were the only mechanism by which the transfer improved knee flexion, however, one would expect that the release of the rectus femoris, without transfer, would be as effective as the rectus femoris transfer. This is not the case. Quantitative gait analysis has demonstrated that the rectus femoris transfer increases knee range of motion more than the release alone (Ounpuu *et al.*, 1993b; Sutherland *et al.*, 1990).

The rectus femoris transfer may be more effective than the release for three reasons: (1) the transferred rectus femoris may develop knee flexion moment during swing and actively contribute to knee flexion; (2) the transfer may better preserve the capacity of the rectus femoris to generate flexion moment about the hip, which could contribute to knee flexion via dynamic coupling [i.e. accelerations of one joint caused by moments about another joint (Zajac and Gordon, 1989)]; (3) transfer of the rectus femoris keeps it from reattaching to the patella, which has been observed during surgeries subsequent to a femoris release. Transferring rectus femoris to any of the four tendon sites may decrease excessive knee extension moment. All transfer sites also prevent reattachment to the patella and preserve moment-generating capacity of the rectus femoris as a hip flexor. However, transferring to semitendinosus, gracilis, or sartorius provides the *possibility* that the transferred rectus femoris could generate knee flexion moment, if it is active during swing. In contrast, rectus femoris transferred to the iliotibial tract does not have the possibility of actively flexing the knee.

Our results suggest that transfer to the semitendinosus may be the most effective means to increase knee flexion moment during swing in patients with isolated swing phase activity of the rectus femoris. Perry (1987) identified four patterns of rectus femoris activity in patients with cerebral palsy. In decreasing frequency of occurrence the EMG patterns identified by Perry are: activity in the swing phase only (58%), continuous and intense activity throughout the gait cycle (20%), strong swing phase activity and a low level of stance phase activity (11%), isolated stance phase activity (11%). For patients with substantial stance phase activity of the rectus femoris, our results suggest transfer to the iliotibial tract. This transfer theoretically eliminates the unwanted knee extension moment during the swing phase and will not produce knee flexion moments during the stance phase, which could exacerbate excessive knee flexion in stance. For patients with swing phase activity of the rectus femoris, our results suggest that transfer to the semitendinosus may be the most effective means to increase knee flexion moment, since it has the greatest flexion moment arm.

Acknowledgements—We gratefully acknowledge the assistance of Kim Statler in the analysis of hip moment arms and Wendy Murray, Dan Alberston, Willie Folie, Joseph Given, Lori Kinzy, Abraham Komattu, Randy Perkins, and Paul Trpkovski for help in the anatomical experiments and data analysis. We are grateful to Sylvia Ounpuu, Luciano Dias, and George Rab for the discussions we have had related to the rectus femoris transfer, and to Idd Delp for help with the illustrations. We thank Musculo Graphics Inc. for donating the software that was used to create and analyze the computer model. This work was supported by the Whitaker Foundation, The Baxter Foundation, and a General Motors Fellowship to Deborah Ringwelski.

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