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Force- and Moment-Generating Capacity of Lower-Extremity Muscles Before and After Tendon Lengthening

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A computer model of the human lower extremity was developed to study how surgical lengthening of tendon affects the force- and moment-generating capacity of the muscles. This model computes the maximum isometric force and the resulting joint moments that each muscle–tendon complex can develop at any body position. Tendon lengthenings were simulated by increasing the tendon length of each muscle–tendon complex and computing the change in the maximum isometric muscle force and joint moments at a specific body position. These simulations showed that the forces and moments developed by the ankle plantarflexors are extremely sensitive to changes in tendon length. For example, at a body position corresponding to the midstance phase of gait, the maximum isometric moment generated by soleus decreased 30% with a 1-cm increase in tendon length, and 85% with a 2-cm increase in tendon length. In contrast, 1- and 2-cm increases in iliopsoas tendon length decreased its hip flexion moment by only 4% and 9%, respectively. This article quantifies the sensitivity of muscle force and joint moments to changes in tendon length for the most commonly lengthened lower-extremity tendons. These results indicate how much each of these tendons should be lengthened to achieve an incremental decrease in muscle force or joint moment.

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Patients with neuromuscular disorders are often treated with tendon transfer and tendon-lengthening surgeries aimed at normalizing joint moments and correcting gait abnormalities. For example, the Achilles tendon is commonly lengthened to correct an equinus deformity (toe walking) in stroke[11] and cerebral palsy patients.[3] Also, the hamstrings may be lengthened in patients who walk with excessive knee flexion, or a crouch gait.[19] Although tendon lengthenings sometimes improve posture and walking, they often compromise the capacity of the muscles to generate force and moments about the joints. When a tendon is lengthened or transferred to a new location, the muscle fibers may be too long or too short to develop active force. Patients who cannot generate sufficient muscle forces and joint moments may be left with weak or dysfunctional legs. For instance, over-lengthening of the Achilles tendon may weaken the plantarflexors and result in excessive dorsiflexion during the stance phase of gait.[3,21] Lengthening the hamstrings may correct excessive knee flexion during stance by decreasing the force in these muscles. Weak hamstrings, however, can lead to genu recurvatum,[2,3,19] inadequate knee flexion during swing,[19] and excessive hip flexion.[7,15]

An understanding of how tendon lengthening affects the force- and moment-generating capacity of each muscle (*i.e.,* muscle strength) is needed to help design effective tendon surgeries. The specific objectives of this study were: (1) to quantify the sensitivity of the maximum isometric force developed by each lower-extremity muscle to changes in tendon length (*i.e.,* to calculate how much force decreases from increasing tendon length), (2) to determine how muscle–fiber length, tendon length, and pennation angle contribute to the sensitivity of muscle force to changes in tendon length, and (3) to quantify the changes in the maximum isometric joint moments that result from increasing tendon length for the most commonly lengthened tendons.

MATERIALS AND METHODS

A computer graphics-based model of the human lower extremity was used to study the mechanical effects of tendon lengthenings. Although this model has been described previously,[6] a summary is given here. The model represents an adult male. The lines of action of 43 muscle–tendon complexes were defined based on their anatomic relationships to three-dimensional bone surfaces. The isometric force-generating property of each muscle–tendon complex was derived by scaling a generic model of muscle and tendon.[26] Four parameters (peak isometric muscle force, optimal muscle–fiber length, tendon slack length, and pennation angle) scale the generic muscle–tendon model to represent a specific muscle–tendon complex.[26] Peak isometric force (based on physiologic cross-sectional area) and optimal muscle–fiber length (the fiber length at which active force peaks) scale the active and passive force–length properties of muscle. Peak isometric force and tendon slack length (the length of tendon beyond which force develops when tendon is stretched) scale the nonlinear force-length property of tendon. Pennation angle specifies the angle between the muscle fibers and the tendon. The kinematics of the lower extremity were defined by modeling the hip, knee, and ankle joints. Thus, the maximum isometric force (*i.e.,* the force developed when a muscle is maximally excited under isometric conditions) and joint moments that each muscle–tendon complex generates can be computed for any body position. The joint moments calculated with the model compare well with isometric joint moments measured during maximum voluntary contractions.[5,6] Because the model assumes full muscle activation and isometric conditions, calculated forces and moments are measures of the maximum force- and moment-generating capacities of the muscles; they are not the forces and moments developed during movement when muscles are, in general, neither fully activated nor isometric.

With the model, the parameters of each muscle–tendon complex can be changed and the effect on the "muscle force curve" observed. The "muscle force curve" is defined here as the maximum isometric force versus joint angle curve. To study the effects of lengthening a particular tendon, the muscle's nominal force curve was plotted. The tendon was then lengthened by a certain amount, and the new muscle force curve was plotted. The change in muscle force that resulted from the increase in tendon length was computed at a specific body position from the muscle force curves. Because muscle force varies with joint angle, the change in force (for a given change in tendon length) depends on body position. Consequently, the changes in the muscle forces were computed at many body positions. For clarity, however, all the results presented in the following section were computed at one body position: slight flexion of the hip, knee, and ankle (Fig. 1). This position was chosen because it occurs in many functional activities (*e.g.,* the stance phase of gait). Also, it was found that the changes in muscle force and joint moments at this body position are similar to those of many other positions.

RESULTS

Computer simulations showed that the decrease in muscle force and joint moments for a given increase in tendon length is *different* for each muscle. Conversely, the increase in tendon length needed to reduce muscle force by a certain amount (*e.g.,* 50%) is different for each muscle (Fig. 2). Notice that there is a wide variation among the muscles in the sensitivity of force to a change in tendon length. For example, soleus force decreased 50% with only a 1.2-cm increase in tendon length. Biceps femoris (long head) and iliopsoas, in contrast, required a 4-cm increase in tendon length to decrease force 50%. In general, the forces developed by the muscles that cross the ankle are more sensitive to a change in tendon length than are the forces generated by the muscles that cross the hip. This variation

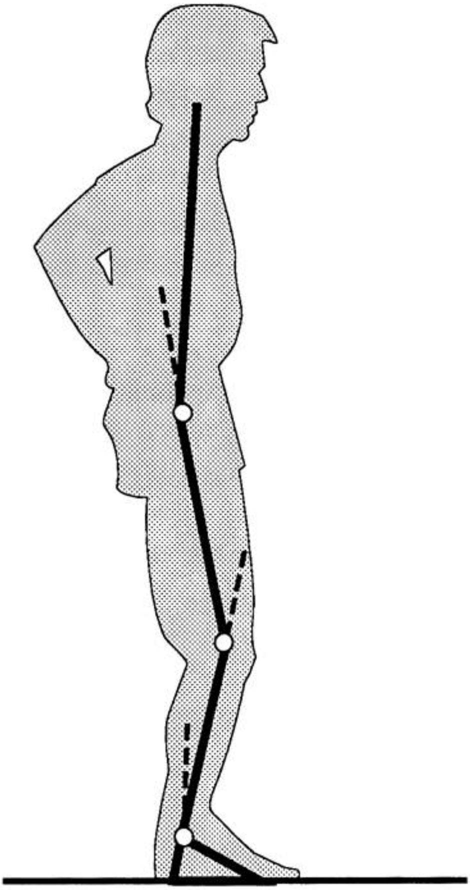

FIG. 1. Body position for which changes in forces and moments are presented. The joint angles are 10° dorsiflexion, 20° knee flexion, and 10° hip flexion. The hip adduction and rotation angles are 0° (anatomic position).

in the sensitivity of the muscle forces to changes in tendon length is caused by differences in the parameters of each muscle–tendon complex. That is, each muscle has a different optimal fiber length, tendon length, and pennation angle, and thus responds differently to tendon lengthening.

EFFECTS OF OPTIMAL FIBER LENGTH

The force developed by a muscle with short fibers is more sensitive to changes in tendon length than the force developed by a muscle with long fibers. A 1-cm increase in tendon length causes an excursion that is significant, in terms of the force–length property, for a muscle with 3-cm fibers (*e.g.*, soleus). In contrast, a 1-cm excursion is insignificant for a muscle with 20-cm fibers (*e.g.*, semitendinosus). Considering the effects of fiber length alone, changing tendon length by 1-cm represents a 33% change in fiber length for the soleus, but only a 5% change for the semitendinosus. Table 1 lists the optimal fiber lengths of the most commonly lengthened muscle–tendon complexes.

Small variations in optimal fiber length can have a large effect on the sensitivity of muscle force to changes in tendon length. To quantify the effect of fiber length, a model of muscle with a fiber length of 5 cm, no pennation, and no tendon (or inelastic tendon) was analyzed. At the joint angle where active force peaks (Θ_o), force decreased 48% for a 2-cm increase in tendon length. Fiber length was then altered and the decrease in force that results from a 2-cm increase in tendon length was computed at Θ_o. Active force decreased 80% when the muscle fibers were 4 cm, and 23% when the fibers were 6 cm. Thus, even a 1-cm change in optimal fiber length has a large effect.

EFFECTS OF TENDON LENGTH

Elastic tendon in series with the muscle fibers increases the range of lengths over which the muscle–tendon complex develops force.[26] This occurs because tendon stretch accounts for part of the muscle–tendon excursion (muscle–tendon excursion = muscle fiber excursion + tendon stretch). Thus, the muscle fibers change length less than the muscle–tendon complex when tendon stretches. Assuming a constant elastic modulus and cross-sectional area of tendon, long tendons stretch more than short tendons for a given force. The magnitude of the effect of tendon stretch on muscle fiber excursion therefore depends on the *ratio* of tendon length to fiber length. Table 1 lists these ratios. Tendon elasticity also decreases the slope of the (active plus passive) muscle force

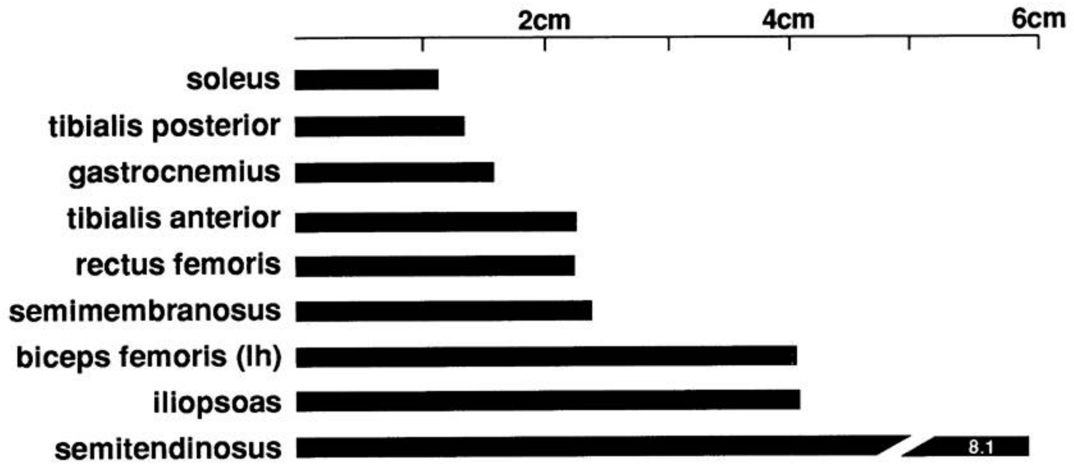

FIG. 2. Change in tendon length needed to reduce muscle force by 50%. The forces in these commonly lengthened tendons were calculated at the body position shown in Figure 1. Tendon lengths were scaled to a 1.8 m adult skeleton.

curve.[26] For example, Figure 3 shows maximum isometric soleus force versus ankle angle with nominal tendon stiffness (elastic) and with tendon stiffness equal to 100 times the nominal value (inelastic).

The variation of tendon lengths among muscles in the lower extremity is about 0.1–30 cm.[9] This leads to a variation in tendon elasticity. To quantify the effect of this variation in tendon elasticity, a model of muscle with 5-cm fibers, 0° pennation, and a 30-cm tendon was derived. With nominal tendon elasticity (3.3% strain at peak active force), active force decreased 29% for a 2-cm increase in tendon length. Without tendon elasticity, active force decreased 48% for a 2-cm

TABLE 1. Parameters for Commonly Lengthened Muscle–Tendon Complexes

Muscle	*Peak Muscle Force*†* *(N)*	*Optimal Fiber Length** *(cm)*	*Pennation Angle** *(degrees)*	*Tendon Slack Length‡* *(cm)*	*Tendon Length/ Fiber Length*
Soleus	2830	3.0	30	26.8	8.9
Tibialis posterior	1270	3.1	12	31.0	10.0
Gastrocnemius§	1600	5.1	14	40.0	7.8
Tibialis anterior	600	9.8	5	22.3	2.2
Rectus femoris	780	8.4	5	34.6	4.0
Semimembranosus	1030	8.0	15	35.9	4.5
Semitendinosus	330	20.1	5	26.2	1.3
Biceps femoris (lh)	720	10.9	0	34.1	3.1
Biceps femoris (sh)	400	17.3	23	10.0	0.6
Iliopsoas¶	800	10.0	8	11.0	1.1

lh, long head; sh, short head.
* Derived from [8,23]; see [5,6] for details.
† Based on physiologic cross-sectional area.
‡ The length of tendon beyond which force develops. This includes the length of free tendon and the length of tendon internal to the muscle belly (aponeurotic tendon). See [6] for details.
§ Peak force is the sum of the two heads; other parameters are the averages of the two heads.
¶ Peak force is the sum of iliacus and psoas; other parameters are the averages of the two muscles.

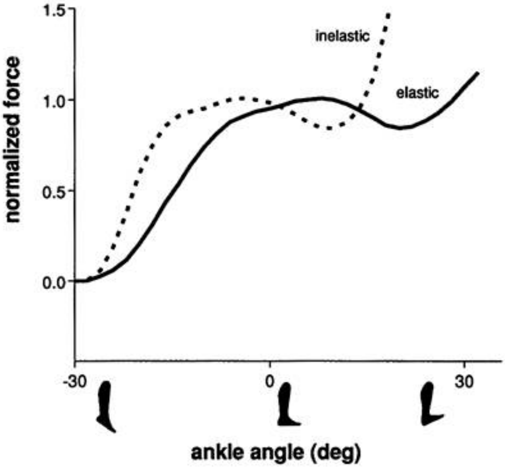

FIG. 3. Active plus passive soleus force versus ankle angle with elastic and inelastic tendon. The solid curve was calculated with nominal tendon elasticity (3.3% tendon strain at peak isometric muscle force). The dotted curve shows the effect of making the tendon inextensible; this also represents a muscle that has a very short tendon or no tendon. Tendon stretch accounts for the difference between the curves. Notice that tendon elasticity tends to decrease the slope of the force versus angle curve. This indicates that tendon compliance decreases the sensitivity of muscle force to changes in muscle–tendon length. Force is normalized by peak isometric active force.

increase in tendon length. Thus, tendon elasticity decreases the sensitivity of muscle force to changes in tendon length.

Effects of Pennation

Pennation decreases the peak force and increases the range of joint motion over which a muscle develops active force.[26] Because muscle force (F^M) is directed along the fibers, the force in tendon (F^T) is decreased by a factor equal to the cosine of the pennation angle (α) (*i.e.*, $F^T = F^M \cos\alpha$).[26] Also, the excursion of the muscle fibers is less than the excursion of the muscle–tendon complex, because the muscle fibers are oriented at an angle to the line of action of the muscle–tendon complex. Thus, pennation tends to decrease the slope of the muscle force versus joint angle curve.

Figure 4 shows the effect of pennation on the soleus force versus ankle angle curve. With 30° of pennation (the nominal pennation angle for the soleus at optimal fiber length),[23] the range of joint angles over which the soleus operates on the ascending region of its force–length curve is greater than it would be if its fibers were unpennated (cf. bars). The effect of pennation is enhanced when the fibers are shorter than optimal length because the pennation angle increases as the fibers shorten. In contrast, the muscle fibers become more aligned with the tendon (*i.e.*, pennation decreases) when the fibers are stretched beyond optimal length. Consequently, pennation has little effect on the passive force characteristics (note the similarity of the slopes of the force curves near full dorsiflexion, where passive force contributes significantly to the active plus passive force in Fig. 4).

The variation in pennation angles among muscles in the lower extremity is about 0°–

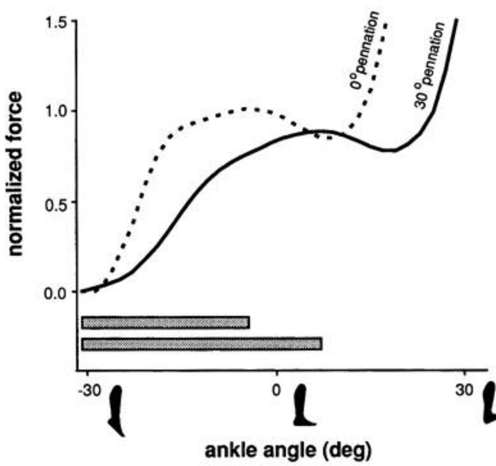

FIG. 4. The effect of pennation on the soleus force versus ankle angle curve. Active plus passive soleus force is plotted for 0° (dotted curve) and 30° (solid curve) of pennation. The bars show the range of ankle angles over which active force increases with muscle length (the ascending region of the force-length curve). Notice that pennation decreases peak force and increases the range of joint motion. Force in both curves is normalized by peak isometric active force for 0° pennation.

30°.[8,23] To assess the effect of this variation in pennation angle on the sensitivity of muscle force to change in tendon length, a model of muscle with 5-cm fibers and an inelastic tendon was analyzed. The decrease in force for a 2-cm increase in tendon length was 48% with 0° pennation, and 30% with 30° pennation. Thus, pennation decreases the sensitivity of muscle force to changes in tendon length.

Relative Importance of Fiber Length, Tendon Length, and Pennation Angle

Optimal muscle–fiber length is the most important parameter in determining the sensitivity of isometric muscle force to changes in tendon length. Optimal fiber lengths vary widely among the muscles in the lower extremity.[8,23] This variation in fiber length is the major factor contributing to the different sensitivities found among the muscles shown in Figure 2.

Tendon elasticity and fiber pennation both decrease the sensitivity of muscle force to change in tendon length. Many muscles have tendons of sufficient length (elasticity) to affect the sensitivity significantly. Of the tendons that are commonly lengthened, the soleus, gastrocnemius, tibialis posterior, peroneus longus, semimembranosus, and biceps femoris (long head) were found to have tendon length to fiber length ratios high enough (greater than 3:11; Table 1) that the slopes of both their active and passive muscle force curves were affected. In contrast, few muscles have pennation angles large enough (more than 20°) to affect sensitivity significantly. The soleus and the biceps femoris (short head) are exceptions, because they have pennation angles of greater than 20° (Table 1).

Effects of Moment Arm

Because some muscles have much smaller moment arms (MAs) than other muscles (*e.g.,* the ankle extension MA of the tibialis posterior is less than one fifth of the Achilles tendon MA),[17] it is important to consider the effects of differing MAs when lengthening tendons. The MA of a muscle affects its moment-generating capacity in two ways. First, since moment about a joint is the product of muscle force and MA, muscles with larger MAs develop greater moments for a given muscle force. Second, MA affects the change in length, or excursion, that a muscle–tendon undergoes as a joint is moved. Moment arm and change in muscle–tendon length ($\partial \ell^{MT}$) are related by the following equation: MA $= \partial \ell^{MT} / \partial \Theta$, where Θ is the joint angle.[1] Thus, muscles with larger MAs exert more moment per unit muscle force, and undergo larger excursions.

At a given body position, the MA of a muscle does not affect the change in muscle force or joint moment that results from tendon lengthening. However, since MA affects excursion, it affects the shape of the muscle–force curve, both before and after tendon lengthening. Figure 5 shows the change in force caused by increasing tendon length for two muscles with the same fiber length, tendon length, and pennation angle, but with different MAs. (A physiologic example of this situation can be demonstrated by comparing the tibialis posterior and the soleus.) Notice that the decrease in force (arrow) is the same for both muscles. However, the force developed by the muscle with the small MA (Fig. 5B) varies less with joint angle than the muscle with a large MA (Fig 5A).

Changes in Joint Moments

The percent change in muscle force for a given change in tendon length depends only on optimal fiber length, tendon length, and pennation angle, as described above. Whether a change in muscle force has a significant effect on the total joint moment, however, depends on the muscle's physiologic cross-sectional area (PCSA) and MA, because PCSA determines peak force,[16] and MA determines moment per unit force. Figures 6, 7, and 8 show the changes in the maximum isometric joint moments calculated for 1- and 2-cm increases in tendon length for

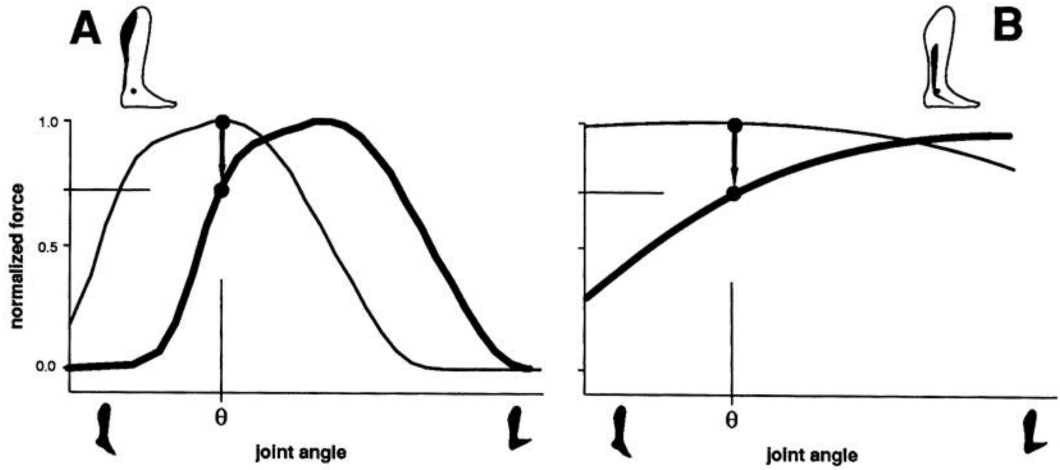

FIG. 5. Force versus joint angle for two muscles with different moment arms. Active force is plotted for muscles with large (A) and small (B) moment arms (MAs). The thin (thick) lines are the muscle force curves before (after) tendon lengthening. Notice that the decrease in force caused by increasing tendon length (arrows) is the same for the two muscles at a specific joint angle (θ). This occurs because the two muscles were given the same fiber length, tendon length, and pennation angle. Thus, the decrease in force at a particular joint angle does not depend on MA. However, since the fibers of the muscle with the large MA undergo a larger excursion as the joint is flexed, the change in force with joint angle is greater in A than B, both before and after tendon lengthening. Force is normalized by peak isometric active force.

the most commonly lengthened lower-extremity tendons.

Figure 6 shows that the plantarflexion moments developed by the soleus, gastrocnemius, and tibialis posterior are extremely sensitive to changes in tendon length. For example, the maximum isometric moment that can be developed by the soleus decreased 30% with a 1-cm increase in tendon length and 85% with a 2-cm increase in tendon length. Because the soleus and the gastrocnemius, together, provide a large percentage (about 90%) of the total plantarflexion moment in this body position, decreasing their

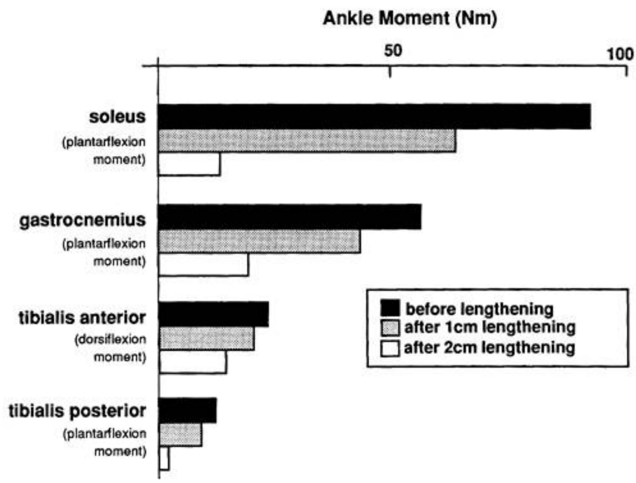

FIG. 6. Maximum isometric ankle moments before and after simulated tendon lengthening. Moments were calculated with 10° dorsiflexion and 20° knee flexion (Fig. 1).

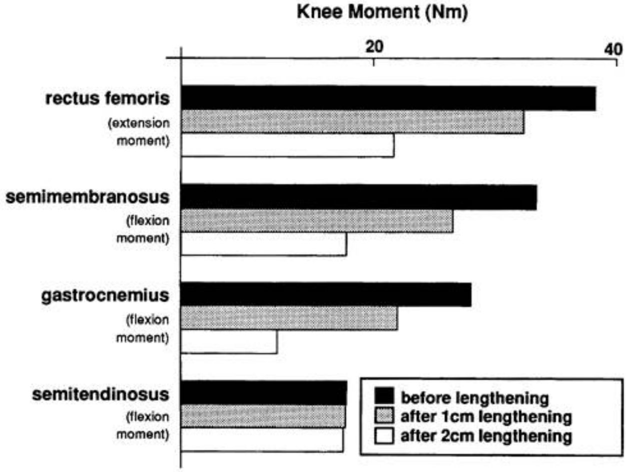

FIG. 7. Maximum isometric knee moments before and after simulated tendon lengthening. Moments were calculated with 10° dorsiflexion, 20° knee flexion, and 10° hip flexion (Fig. 1).

moment-generating capacity has a significant effect on the total plantarflexion strength. Tibialis posterior moment also decreases substantially with a change in tendon length. This has a small effect on total plantarflexion moment, however, because the tibialis posterior contributes only a small percentage to the total plantarflexion moment because of its small moment arm.

Of the ankle muscles, the moment developed by the tibialis anterior is the least sensitive to change in tendon length. The tibialis anterior is less sensitive because it has long fibers (9.8 cm) and a long tendon (22 cm), both of which decrease sensitivity.

Figure 7 shows the changes in the knee moments for 1- and 2-cm increases in tendon length. The knee extension moment developed by the rectus femoris decreased 17% and 48% for 1-cm and 2-cm increases in tendon length, respectively. The knee flexion moments generated by the semimembranosus and the gastrocnemius are both fairly sensitive to a change in tendon length. The knee flexion moment generated by the semitendinosus, however, decreases only slightly with an increase in tendon length. Because the semimembranosus is more sensitive than the semitendinosus, and provides a greater percentage of the total knee flexion moment, a

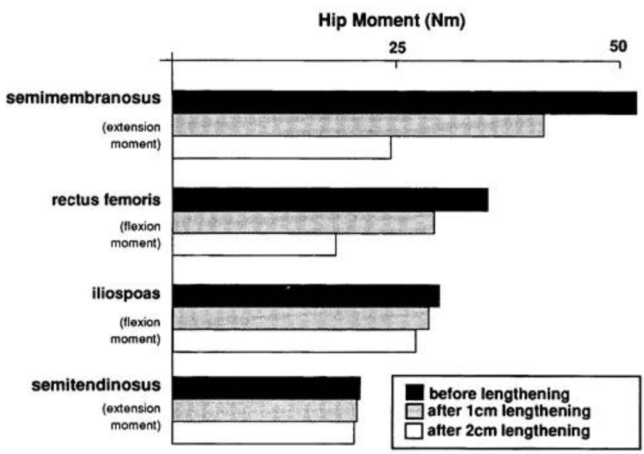

FIG. 8. Maximum isometric hip moments before and after simulated tendon lengthening. Moments were calculated with 20° knee flexion and 10° hip flexion (Fig. 1).

much greater decrease in total knee flexion moment results from lengthening the semimembranosus.

When the tendon of a biarticular muscle is lengthened, the effects on the moments about both spanned joints must be considered. For example, when the hamstrings are lengthened to correct excessive knee flexion during stance (*i.e.,* "crouch gait"), the extension moment they exert about the hip is also changed. Figure 8 shows the change in hip moment for 1- and 2-cm lengthening of the hamstrings and other tendons. For the hamstrings, the semimembranosus is capable of generating more than twice as much extension moment as the semitendinosus because it has a larger PCSA.[8,23] Also, the moment developed by the semimembranosus is much more sensitive to changes in tendon length because it has shorter fibers. Therefore, the hip extension moment (*i.e.,* hip extension strength) is decreased significantly by lengthening the semimembranosus but is changed very little by lengthening the semitendinosus.

For the hip flexors, it was found that the maximum isometric moment developed by the rectus femoris is more sensitive to change in tendon length than the moment developed by the iliopsoas (Fig. 8). Also, the rectus femoris can generate slightly more hip flexion moment than the iliopsoas. Although the iliopsoas has a larger PCSA,[8,23] it has less potential to generate flexion moment than the rectus femoris because it has a smaller MA. Thus, the rectus femoris is a slightly stronger hip flexor than the iliopsoas and is more sensitive to a change in tendon length.

DISCUSSION

Each muscle has a unique set of muscle–tendon parameters (physiologic cross-sectional area, optimal fiber length, pennation angle, and tendon length) that determine its isometric force-generating characteristics. Furthermore, each muscle has a unique three-dimensional geometric relationship with respect to the joint or joints it spans. The interplay of the muscle–tendon parameters and the musculoskeletal geometry determines the capacity of each muscle to generate moment about the joints. This study used a model that specifies the isometric force- and moment-generating capacity for all the major muscles in the lower extremity to simulate the biomechanical consequences of tendon lengthenings. The simulations indicate that the forces developed by some muscles (particularly the ankle plantarflexors) are sensitive to changes in tendon length. Thus, small changes in tendon length result in large changes in muscle force. By contrast, other muscles (*e.g.,* the iliopsoas and the semitendinosus) are far less sensitive to change in tendon length and must therefore be lengthened more aggressively to achieve a significant decrease in force. The simulation results quantify the sensitivity of muscle force and joint moment to changes in tendon length for the most commonly lengthened lower-extremity tendons.

Confidence in Muscle–Tendon Parameters

The confidence in the simulation results are limited by the accuracy of the muscle–tendon parameters. Estimates of muscle–fiber length are particularly important. Three independent investigations of lower-extremity muscle architecture have reported remarkably consistent fiber lengths.[8,22,23] Thus, the authors are confident that the fiber lengths used in this study[8,23] are reasonable.

However, the difference between fiber length and fascicular length, and complex muscle architectures, must be considered. Wickiewicz *et al.*[23] dissected bundles of ten to 20 muscle fibers (called muscle fascicles) and measured fascicular length. Friederich and Brand[8] measured both fascicular and fiber lengths. For many muscles, fascicular lengths and fiber lengths are similar.[8] Based on these data, the model assumes that fascicular lengths and fiber lengths are equal. Other data, however, demonstrate that the fibers of some parallel-fibered muscles in the cat hindlimb do not run from internal tendon to internal tendon, as the fascicles do, and thus are shorter than the muscle fascicles.[10] If human parallel-fibered muscles (*e.g.,* the semitendinosus) have fibers shorter than the fascicular lengths used in this study, then the effect of a change in tendon length on force and moment for those muscles would be greater than the effects reported here.

The semitendinosus muscle has another architectural specialization related to fiber length. There is a tendinous septum near the middle of the muscle belly that separates it into two distinct compartments. Because these compartments are in series, the semitendinosus was treated as a single muscle with a fiber length equal to the sum of the fiber lengths of the two compartments.[23] In effect, therefore, it was assumed that both compartments are simultaneously rather than differentially excited during motor tasks.

Measured values for PCSA, which scale peak muscle force,[16] are less consistent than measured fiber lengths. Friederich and Brand[8] found that "normalized" PCSA values (*i.e.,* PCSA of a muscle/average PCSA of all muscles) measured in six cadavers were somewhat consistent (average standard deviation for all the muscles was 29%).[8] Normalized PCSA measured by Brand *et al.*[4] in 15 anatomic specimen arms had an average standard deviation of 26% (calculated from mass fraction in Table 1 of the study by Brand *et al.*). Because PCSA does not affect the percentage change in force for a given change in tendon length, variability in this parameter does not affect the *relative* sensitivities (*i.e.,* the data shown in Fig. 2 are not affected). Because PCSA does affect the magnitude of muscle force, however, the *absolute* moment-generating capacities shown in Figures 6–8 would be affected if the PCSA of the muscles were to vary significantly among subjects. Nonetheless, the relationships between the moments shown in Figures 6–8 would not be affected if the *normalized* PCSA of the muscles were intersubject independent.

Measurements of pennation angle are very consistent.[8,23] The values used in this study are therefore adequate, especially because small changes in pennation angle have little effect on the sensitivity of muscle force to changes in tendon length.

Because no experimental measurements of tendon slack length have been reported, a study was undertaken to assess the accuracy of the estimates of tendon length used here.[6] This study showed that the estimates of tendon length for muscles with high ratios of tendon length to fiber length (*e.g.,* the soleus, the tibialis posterior, and the gastrocnemius) are reliable, because tendon lengths had to be specified accurately so that calculated joint moments matched experimental joint moments.[5] Estimates of tendon length for muscles with small ratios of tendon length to fiber length (*e.g.,* the iliopsoas and the semitendinosus) are less reliable. However, since force is less sensitive to changes in tendon length for these muscles, accurate estimates are less critical because an error in tendon length has a smaller effect on the muscle–force curve.

Assumptions and Limitations

It is important to discuss the assumptions and limitations of using the present computer model to study the mechanics of tendon lengthenings. First, and most importantly, the simulation results do not account for the effects of muscle–tendon adaptation that can accompany immobilization after tendon surgery. Immobilization can decrease the peak force of a muscle,[24] alter the number of sarcomeres in a muscle fiber,[18,24] and change the elasticity of tendon.[25] In the model, the peak force, fiber length, and tendon elasticity of each muscle–tendon complex can be altered to simulate the mechanical effects of muscle–tendon adaptation. However, these parameters were kept constant in Figures 2, 6, 7, and 8, however. These figures simply quantify the variation in the sensitivity among the muscles that results from normal differences in the muscle–tendon parameters.

The differences in the muscle–tendon parameters between muscles is much larger than the changes in these parameters that could result from adaptation of an individual muscle. For example, the variations of fiber length among muscles in the lower extremity can be as large as 1000%,[8,23] but optimal muscle–fiber lengths have been shown to increase only about 20% when muscle is held in an elongated position.[18] Thus, muscle–tendon adaptation would not affect the relative sensitivities presented in Figure 2.

This is not to say that adaptation is unimportant in determining the biomechanical consequences of tendon lengthenings. Adaptation is in fact extremely important, especially for muscles that are sensitive to a parameter change. For instance, the model indicates that even a 1-cm decrease in the fiber length of the soleus would decrease the range of joint angles over which active force is generated from 60° to 40° and alter the force developed at each joint angle. Muscles that are insensitive to a parameter change must adapt much more to have a substantial effect on the force-generating characteristics.

Truscelli *et al.*[20] postulated that the lack of muscle growth during bone growth is often the cause of equinus in children with cerebral palsy. The model used here does not account for the effects that may accompany skeletal growth. Figures 2, 6, 7, and 8 should therefore be used to understand the acute changes in the muscle forces and joint moments that result from lengthening tendon, not the changes that may result after growth and adaptation.

Another limitation of the computer model is that it represents the musculoskeletal system of a healthy person of nominal stature (1.8 m). Tendon lengthenings are frequently performed on stroke and cerebral palsy patients, however, whose condition may be complicated by muscle–tendon contracture. How, then, can this model be used to gain insight into surgeries performed on patients whose muscle–tendon parameters may be abnormal? If experimental data were available to indicate how the muscle–tendon parameters are affected by central nervous system pathology such as stroke or cerebral palsy, the parameters of the model could be modified to represent these pathologic conditions. Unfortunately, no such experimental data exist. The nominal model, however, can be used to study how pathologic changes would affect the sensitivity of the muscle forces. For example, if the muscle fibers in cerebral palsy patients are shorter than the average values used in this study, then the changes in muscle force that result from tendon lengthening would be even greater than the changes reported here. Also, analysis of the nominal model suggests that muscle–fiber length is the most important parameter in determining the sensitivity of muscle force to changes in tendon length. Future experiments therefore should investigate how fiber length is affected in pathologic states.

Finally, even if one knew precisely how the musculoskeletal system changes with tendon lengthening, the functional result of these surgeries would still be somewhat unpredictable because of abnormal muscle activation patterns that often accompany central nervous system pathology.[13,14] This study focused on the force- and moment-generating capacities of the muscles because they are affected directly by tendon surgery. The current model, however, cannot be used to study more complex issues, such as how tendon lengthening indirectly affects force production through its influence on neural control patterns.

Clinical Implications

The results presented here have practical implications with regard to tendon lengthenings. It was found that the forces and moments developed by the soleus and gastrocnemius change significantly with small changes in tendon length. This suggests that the Achilles tendon should be lengthened conservatively to avoid plantarflexion weakness. Clinical studies also suggest conservative lengthening of the Achilles tendon.[3] The observation that lengthening of the Achilles tendon may weaken the soleus more than the gastrocnemius supports those who recommend isolated gastrocnemius lengthening as a means to control plantarflexion weakness in cerebral palsy patients.[12] In contrast to the soleus, the force and moment developed by the iliopsoas is relatively insensitive to changes in tendon length. This indicates that it can be lengthened more aggressively without much decrease in force. The difference in the fiber lengths of the muscles of the hamstrings has similar implications for surgeons who lengthen multiple hamstrings.

The results presented here can also be applied to understand the force- and moment-generating capacity of muscles used in tendon transfers. If the force developed by a muscle is sensitive to changes in tendon length, then it is also sensitive to changes in origin-to-insertion length. That is, if the maximum isometric force of a muscle decreases a certain amount from lengthening its tendon 2 cm, then its force would decrease by the same amount if a tendon transfer decreased its origin-to-insertion length 2 cm. Thus, the sensitivity results presented above apply to both tendon lengthenings and tendon transfers.

There are practical implications of this connection between tendon transfers and tendon lengthenings. If a muscle that is sensitive to length change (*e.g.,* the tibialis posterior) is to generate active force after a transfer, the transfer must be performed such that the muscle fibers are near optimal length. This may be difficult to accomplish for a muscle with short fibers. Conversely, muscles that are less sensitive to length change (*e.g.,* the rectus femoris) are more likely to generate active force after a transfer, even if the origin-to-insertion length is changed significantly.

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