



Sex comparison of hamstring structural and material properties

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ABSTRACT

Background: Musculotendinous stiffness provides an estimate of resistance to joint perturbation, thus contributing to joint stability. Females demonstrate lesser hamstring stiffness than males, potentially contributing to the sex discrepancy in anterior cruciate ligament injury risk. However, it is unclear if the sex difference in hamstring stiffness is due to differences in muscle size or to inherent/material properties of the musculotendinous unit. It was hypothesized that hamstring stiffness, stress, strain, and elastic modulus would be greater in males than in females, and that hamstring stiffness would be positively correlated with muscle size.

Methods: Stiffness was assessed in 20 males and 20 females from the damping effect imposed by the hamstrings on oscillatory knee flexion/extension following joint perturbation. Hamstring length and change in length were estimated via motion capture, and hamstring cross-sectional area was estimated using ultrasound imaging. These characteristics were used to calculate hamstring material properties (i.e., stress, strain, and elastic modulus).

Findings: Stiffness was significantly greater in males than in females ($P < 0.001$). However, stress, strain, and elastic modulus did not differ across sex ($P > 0.05$). Stiffness was significantly correlated with cross-sectional area ($r = 0.395$, $P = 0.039$) and the linear combination of cross-sectional area and resting length ($R^2 = 0.156$, $P = 0.043$).

Interpretation: Male's hamstrings possess a greater capacity for resisting changes in length imposed via joint perturbation from a structural perspective, but this property is similar across sex from a material perspective. Females demonstrate lesser hamstring stiffness compared to males in response to standardized loading conditions, indicating a compromised ability to resist changes in length associated with joint perturbation, and potentially contributing to the higher female ACL injury risk. However, the difference in hamstring stiffness is attributable in large part to differences in muscle size.

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1. Introduction

Anterior cruciate ligament (ACL) injury influences a substantial number of individuals, with females displaying a considerably greater injury risk compared to males (Arendt et al., 1999). Non-contact ACL injury typically occurs immediately following ground contact during landing and gait activities (Griffin et al., 2006). The quadriceps muscles become active prior to ground contact during these activities and remain active throughout the loading phase (Fagenbaum and Darling, 2003; Colby et al., 2000) in an attempt to attenuate landing forces and control downward acceleration of the body. Quadriceps contraction loads the ACL by producing anterior tibial shear force and translation (Li et al., 1999; Withrow et al., 2006a), and isolated quadriceps force is capable of rupturing the cadaveric ACL (DeMorat et al., 2004). Therefore, any factor which limits anterior tibial translation and shear force would theoretically limit ACL loading and injury risk.

The hamstrings are capable of limiting the load placed on the ACL by resisting anterior tibial translation and shear force. The addition of hamstring force in the cadaveric knee reduces anterior tibial translation (Li et al., 1999), anterior tibial shear force (MacWilliams et al., 1999), ACL strain (Withrow et al., 2006b), and ACL loading (Li et al., 1999), and similar reductions in ACL strain have been reported *in vivo* during knee extension exercises (Beynon et al., 1995). Interestingly, heightened hamstring activity during weight bearing (Andriacchi and Birac, 1993) and heightened hamstring stiffness (Jennings and Seedhom, 1998) have been noted in ACL-deficient individuals. These adaptive changes in hamstring properties may enhance knee joint stability in an attempt to compensate for the absence of the ACL.

By virtue of their insertions on the posterior shank, the hamstrings are lengthened via anterior tibial translation. The hamstrings respond to this imposed lengthening with an increase in tensile force which resists further lengthening. Musculotendinous stiffness (MTS) refers to the ratio of the change in tensile force to the change in musculotendinous length associated with joint motion ($\Delta \text{Force}/\Delta \text{Length}$). Heightened MTS may enhance joint

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stability in that for a given change in length, a stiffer musculotendinous unit will respond with a greater increase in resistive tensile force. By responding with a larger resistive tensile force, stiffer hamstrings may be able to limit anterior tibial translation and shear force, and reduce the force imparted to the ACL.

Previous literature indicates that hamstring MTS is lesser in females than in males (Granata et al., 2002; Blackburn et al., 2004). Given the potential contributions of hamstring MTS to knee joint stability and ACL injury risk, this sex discrepancy may contribute to the greater incidence of ACL injury in females. However, MTS is a structural property, thus the greater hamstring MTS noted previously in males is at least partially attributable to greater hamstring size. While stiffness is a structural property reflecting the geometry of the musculotendinous unit, elastic modulus (i.e., the ratio of stress to strain) is the material equivalent of stiffness, thus it represents stiffness independent of muscle size. Previous investigations have demonstrated that both MTS and elastic modulus of the triceps surae (Blackburn et al., 2006) and Achilles tendon (Kubo et al., 2003) are greater in males than in females. It is currently unclear if the previously noted sex difference in hamstring MTS is attributable to discrepancies in hamstring size or inherent (material) properties of the musculotendinous unit. Therefore, the purposes of this investigation were to replicate the previous finding of greater hamstring MTS in males and to evaluate the dependency of this sex difference on hamstring geometry. It was hypothesized that hamstring MTS, stress, strain, and elastic modulus would be greater in males than in females, and that MTS would be positively correlated with muscle size.

2. Methods

2.1. Subjects

Forty healthy, physically active individuals (20 males, 20 females) volunteered as subjects for this investigation. To be eligible for participation, subjects were required to be free of lower extremity musculoskeletal injury within the 6 months prior to data collection, and have no history of ACL injury, knee surgery, or neurological disorder. All subjects read and signed an approved informed consent document prior to participation. Subject demographic data are presented in Table 1. As hamstring MTS is similar across limbs (Jennings and Seedhom, 1998), all data were collected from the right limb only.

2.2. Experimental procedures

Two separate assessments were conducted in a counterbalanced order: hamstring MTS and ultrasonic imaging of the hamstrings. Hamstring MTS was estimated by modeling the knee as a single-degree-of-freedom mass–spring system and evaluating the damping effect imposed by the hamstrings on oscillatory knee flexion/extension as described by Blackburn et al. (2004). Subjects were positioned prone with the thigh supported in 30° of flexion below horizontal, and a load representing 10% body mass was secured to the distal shank. The investigator passively positioned the shank on the horizontal, and the subject was required to contract the hamstrings to support the applied load in this position. Given the orientation of the thigh, this isometric contraction placed the knee in approximately 30° of flexion (Fig. 1).

Table 1
Subject demographics: mean (SD).

	Height (cm)	Mass (kg)	Age (yr)
Males (<i>n</i> = 20)	179.3 (8.8)	75.9 (10.9)	20.7 (1.6)
Females (<i>n</i> = 20)	164.0 (5.6)	61.4 (9.2)	20.4 (1.5)

At a random point within 5 s of hamstring contraction, the investigator applied a brief downward manual perturbation approximately at the calcaneus, initiating oscillatory knee flexion/extension. An accelerometer (PCB Piezotronics, Depew, NY, USA) was fixed to a rigid splint secured on the posterior shank and foot, allowing measurement of shank segment tangential acceleration (Fig. 2). With knowledge of the time instances of the first two oscillatory peaks in the tangential acceleration profile (t_1 and t_2), the damped frequency of oscillation ($\frac{1}{t_2 - t_1}$) was derived and substituted in the equation

$$K = 4\pi^2 mf^2 \quad (1)$$

where K is linear stiffness (N/cm), m is the summed mass of the shank and foot segment (Winter, 1990) and the applied load, and f is the damped frequency of oscillation. Subjects performed five trials with 1 min of rest provided between trials to reduce the likelihood of fatigue. This method has been demonstrated to produce moderate-to-high intra-session reliability across repeated measurements ($ICC_{2,1} = 0.70$; $SEM = 1.63$ N/cm) (Blackburn et al., 2004).

To derive the material properties stress, strain, and elastic modulus, it was necessary to measure hamstring length and cross-sectional area (CSA). Modeling the knee joint as a single-degree-of-freedom system assumes that only flexion and extension occur during the perturbation, and that axial (internal/external) rotation at the tibiofemoral joint is negligible. Within this model, the contributions of the medial (semimembranosus and semitendinosus) and lateral (biceps femoris) hamstrings to knee flexion moment are assumed to be equal (Kellis et al., 2005), as the sum of the axial rotation moments must be zero (i.e., rotational equilibrium). Therefore, the length and CSA of the biceps femoris long head (i.e., the lateral hamstrings) were used to represent the corresponding values for the hamstrings group. These values were then used to estimate stress and strain, which were in turn used to estimate elastic modulus.

Biceps femoris length was measured using an electromagnetic motion capture system (MiniBirds, Ascension Technology Corp., Burlington, VT, USA). Electromagnetic sensors were secured on the pelvis, thigh, and shank, and a segment-linkage model of the lower extremity was created by digitizing the knee, ankle, and hip joint centers using The Motion Monitor software (Innovative Sports Training, Chicago, IL, USA). Additionally, the origin (ischial tuberosity) and insertion (fibular head) of the biceps femoris long head were digitized, and hamstring length was calculated as the linear distance between these three-dimensional coordinates. These kinematic data were sampled during the hamstring MTS assessment, and were used to estimate the change in hamstring length induced by the perturbation and the resting/pre-perturbation length. While this procedure ignores the curvature/wrapping

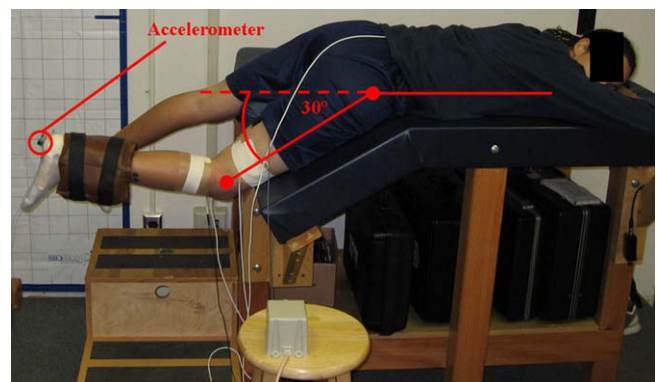


Fig. 1. Subject positioning during hamstring MTS assessments.

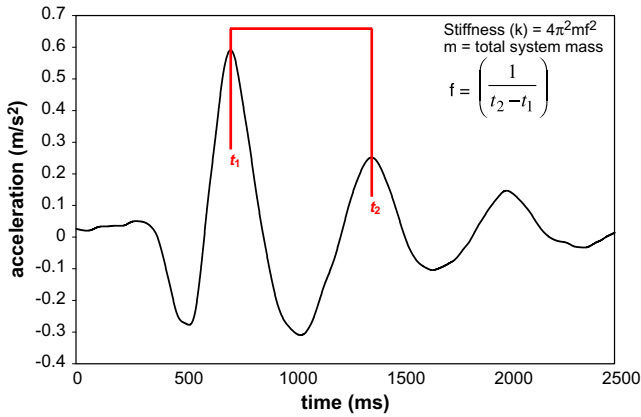


Fig. 2. Tangential acceleration of the shank segment during stiffness assessments.

of the musculotendinous unit attributable to connective tissue and likely underestimates the true muscle length (Arnold and Delp, 2001), the magnitude of this difference is likely minimal.

Biceps femoris anatomical CSA was measured using ultrasound imaging (Fig. 3). The anatomical CSA was chosen over the physiological CSA to facilitate comparison to published values obtained using magnetic resonance imaging (Magnusson et al., 1997). Ultrasound imaging provides estimates of muscle CSA which are reliable across repeated scans (ICC = 0.997–0.999), and are valid relative to magnetic resonance imaging (ICC = 0.998–0.999) (Reeves et al., 2004). Subjects were positioned prone, and B-mode ultrasound (LOGIQe, General Electric, Milwaukee, WI, USA) was applied to the posterior thigh using a 7 MHz linear-array transducer to acquire axial plane images at the midpoint of the thigh, 50% of the distance between the greater trochanter and the lateral knee joint line. A silicone template was secured over the posterior thigh perpendicular to the longitudinal axis, and the transducer was placed within the template to maintain its location and orientation in the axial plane. In all subjects, the width of the biceps femoris long head was greater than that of the transducer (~4 cm), thus it was necessary to capture two images to encompass the entire muscle. To accommodate this limitation, a hyperechoic marker (23-gauge hypodermic needle) which was easily identifiable in

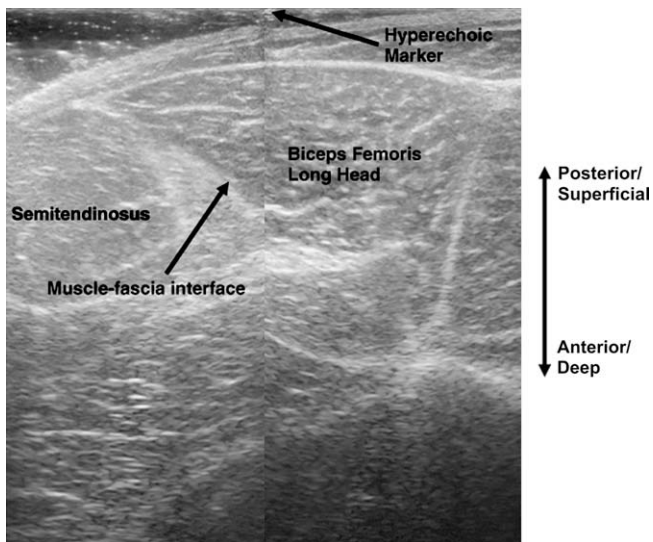


Fig. 3. Axial plane ultrasound image of the biceps femoris long head captured at the midpoint of the thigh. The hyperechoic marker can be seen in the top left corner of the image on the right. Image depth is 8 cm.

the image was embedded in the silicone template. In the first image, the lateral portion of the muscle was scanned with the hyperechoic marker visible in the most posteromedial (top left) margin of the field of view. The medial portion of the muscle was scanned in the second image with the marker visible in the most posterolateral (top right) margin of the field of view. With the first image displayed on half of the computer screen, the transducer was moved medially until the marker was no longer visible. This procedure essentially overlapped the two images relative to the external marker, thus allowing for a combined image which encompassed the entirety of the muscle (Reeves et al., 2004). CSA measurements were made using a public domain image processing and analysis program (Image J, National Institutes of Health, USA). This software platform was used to trace the deep/internal interface between the hyperechoic epimysium (external fascia) and the hypoechoic muscle tissue, and to calculate the area of the resulting polygon.

The change in force during the stiffness assessment was determined by multiplying the stiffness value (N/cm) by the change in length of the biceps femoris (cm).

$$\Delta F_{ham} = \frac{\Delta F}{\Delta \ell} \times \Delta \ell_{BF} \quad (2)$$

Assuming equal force contributions from the medial and lateral hamstrings (Kellis et al., 2005), ΔF_{ham} was halved to derive the change in force of the biceps femoris.

$$\Delta F_{BF} = \frac{\Delta F_{ham}}{2} \quad (3)$$

Next, ΔF_{BF} was divided by the biceps femoris CSA to produce biceps femoris stress (σ).

$$\sigma_{BF} = \frac{\Delta F_{BF}}{CSA_{BF}} \quad (4)$$

Biceps femoris strain (ϵ) was then calculated by dividing the change in length induced by the perturbation ($\Delta \ell_{BF}$) by the initial, pre-perturbation length of the biceps femoris (ℓ_{0BF}).

$$\epsilon_{BF} = \left(\frac{\Delta \ell_{BF}}{\ell_{0BF}} \right) \times 100 \quad (5)$$

Finally, biceps femoris elastic modulus (E) was estimated by calculating the ratio of stress to strain.

$$E = \frac{\sigma_{BF}}{\epsilon_{BF}} \quad (6)$$

2.3. Data sampling, reduction, and analysis

Electromagnetic sensor data were sampled at 100 Hz while accelerometer data were sampled at 1000 Hz. Electromagnetic sensor data were time-synchronized to the accelerometer data and re-sampled to 1000 Hz. All data were lowpass filtered at 10 Hz (4th order zero-phase-lag Butterworth). A custom computer program (LabVIEW, National Instruments, Austin, TX, USA) was used to calculate MTS, biceps femoris length, stress, strain, and elastic modulus.

Hamstring MTS, stress, strain, elastic modulus, CSA, and resting length were compared across sex using one-tailed independent-samples t -tests. Musculotendinous structures are viscoelastic in nature, thus the force response to the imposed lengthening is loading-rate sensitive. To validate that the magnitude of the perturbation used to induce oscillatory knee flexion/extension was similar between males and females (i.e., standardized loading conditions), we compared the perturbation force across sex using a two-tailed independent-samples t -test. This value was calculated as the product of the peak tangential shank segment acceleration and system mass (i.e., summed mass of the shank and foot segment (Winter,

1990) and the applied load). Simple linear regression was used to assess the relationships between hamstring MTS (criterion variable) and CSA and resting length, respectively. Additionally, stepwise multiple regression was used to evaluate the relationship between hamstring MTS and the linear combination of CSA and resting length. CSA was entered first as the lone predictor variable followed by resting length. Statistical analyses were conducted using commercially available statistical software (SPSS Inc., Chicago, IL, USA), with statistical significance established *a priori* as $\alpha = 0.05$.

3. Results

Hamstring MTS was significantly greater in males than in females ($P < 0.001$). The perturbation force did not differ across sex [mean (SD): males = 30.77 (10.96) N; females = 27.31 (9.02) N; $P = 0.408$]. In general, the hamstrings were larger in males than in females as indicated by significant differences in both resting length ($P = 0.003$) and CSA ($P < 0.001$) (Table 2). When these geometric discrepancies were accounted for, however, hamstring stress ($P = 0.486$), strain ($P = 0.338$), and elastic modulus ($P = 0.230$) did not differ significantly across sex (Table 3).

While MTS was significantly and positively related to CSA ($r = 0.395$, $P = 0.039$), the relationship between MTS and resting length was non-significant ($r = 0.012$, $P = 0.813$). Stepwise multiple regression indicated that the linear combination of CSA and resting length (full model) was significantly correlated with MTS ($R^2 = 0.156$, $P = 0.043$), but that the addition of resting length to the model as the second predictor variable did not significantly increase the amount of explained variance in MTS (R^2 -change < 0.001 , $P = 0.971$).

4. Discussion

The primary findings of this investigation were that while hamstring MTS was greater in males than in females, elastic modulus did not differ significantly across sex. Additionally, hamstring muscle size predicted ~16% of the variance in hamstring MTS. These findings indicate that males' hamstrings possess a greater capacity for resisting changes in length imposed via joint motion from a structural perspective, but that this property is at least partially attributable to greater muscle size in males, and is similar across sex from a material perspective.

The finding of greater hamstring MTS in males and the reported MTS values are in agreement with previous investigations by Blackburn et al. (2004) and Granata et al. (2002). Additionally, both of these investigations noted a dependency of MTS on anthropometric factors under the assumption that muscle size is proportional to height and mass. Specifically, Blackburn et al. standardized MTS in angular units (N m/rad) to the applied moment (product of system mass and shank segment length; N m) and found that when between-subject anthropometric discrepancies were accounted for, the sex difference in hamstring MTS was non-significant. Granata et al. reported a positive relationship between MTS and the applied moment, indicating that MTS increased as a function of height and mass.

Table 2
Hamstring geometry characteristics: mean (SD).

	Males (n = 20)	Females (n = 20)	P-value
CSA (cm ²)	11.9 (3.0)	8.2 (1.3)	<0.001*
Resting length (cm)	44.8 (3.7)	41.4 (2.9)	0.003*

* Significant difference across sex.

Table 3
Experimental data: mean (SD).

	Males (n = 20)	Females (n = 20)	P-value
MTS (N/cm)	14.0 (3.1)	10.2 (1.9)	<0.001*
Stress (KPa)	1304.1 (1049.5)	1292.0 (1140.0)	0.486
Strain (% ϵ_{0BF})	0.5 (0.3)	0.6 (0.4)	0.338
Elastic modulus (KPa)	2811.8 (883.3)	2634.1 (594.3)	0.230

* Significant difference across sex.

Our values for biceps femoris long head CSA are slightly less than those reported in the literature. Magnusson et al. (1997) measured biceps femoris long head CSA in 14 males using magnetic resonance imaging, reporting a mean (SD) value of 13.99 (1.89) cm². In comparison, the mean value for male subjects in the current investigation was 11.87 (3.01) cm². The discrepancy in these values is likely attributable to differences in the subject samples, as Magnusson et al. investigated elite athletes, while we investigated individuals who met a minimum physical activity requirement. Additionally, our estimates of biceps femoris long head length based on three-dimensional coordinates of the digitized ischial tuberosity and fibular head are in agreement with similar values reported in cadaveric specimens (Woodley and Mercer, 2005).

Our primary hypothesis was that the greater hamstring MTS noted in males would persist after accounting for differences in muscle size (i.e., estimation of elastic modulus). This assumption was made based on previous literature indicating that both MTS and elastic modulus are greater in males in the triceps surae muscle complex (Blackburn et al., 2006; Kubo et al., 2003), findings which imply sex differences in material properties of the musculotendinous unit. However, our data indicate that while structural properties of hamstrings differ across sex (i.e., MTS), the material properties (i.e., stress, strain, and elastic modulus) do not. The discrepancy underlying these investigations is likely due to differences in architectural characteristics of the triceps surae and hamstrings which contribute to force production.

The literature indicates in both cadaveric and *in vivo* imaging investigations that the muscle fiber length of the biceps femoris long head is greater than those of the triceps surae components (i.e., medial and lateral gastrocnemius and soleus) (Wickiewicz et al., 1983; Woodley and Mercer, 2005). However, the angle of pennation and physiological CSA (PSCA) of the biceps femoris long head are substantially smaller than those of the triceps surae components (Wickiewicz et al., 1983; Woodley and Mercer, 2005). Classic characterization of muscle function suggests that muscles composed of relatively long fibers (i.e., a large number of sarcomeres arranged in series) with small pennation angles are best suited for joint excursion, while those composed of short fibers (i.e., large number of fibers arranged in parallel) typically possess relatively larger angles of pennation and PCSAs, and are best suited for force production (Wickiewicz et al., 1983; Woodley and Mercer, 2005). Additionally, muscle force production is proportional to PCSA (Fukunaga et al., 1996). These architectural characteristics suggest that the triceps surae is better suited for force production, whereas the hamstrings are best suited for joint excursion.

Sex differences in pennation angle (Chow et al., 2000; Kubo et al., 2003) and muscle fiber length (Chow et al., 2000) have been reported in the triceps surae, with females possessing lesser angles of pennation and longer muscle fibers. These discrepancies in muscle architecture would persist after accounting for CSA and musculotendinous length. Therefore, the triceps surae muscle complex appears to differ architecturally across sex, thus contributing to sex differences in material properties of the triceps surae. We were unable to identify any previous investigations which identified similar sex differences in the hamstrings. However, the hamstrings

are typically classified as fusiform in structure, with the majority of fibers arranged essentially in parallel with the longitudinal axis. Cadaveric estimates of biceps femoris long head pennation angle range from 0° to 7° (Friederich and Brand, 1990; Wickiewicz et al., 1983), thus it seems unlikely that variations within this limited range would differ systematically across sex in a manner which would have an appreciable influence on muscle force production and MTS. These discrepancies may explain why elastic modulus differs in the triceps surae, yet not in the biceps femoris. Muscle fiber type also influences force production. However, we are unaware of any investigations detailing sex differences in fiber type distribution in the hamstrings, and previous literature regarding sex differences in fiber distribution in other muscles is inconsistent (Staron et al., 2000; Simoneau and Bouchard, 1989). As such, it is unlikely that a systematic sex difference in fiber type distribution contributed to our results (Kubo et al., 2003). However, it is important to note that while controlling for muscle geometry nullified the sex difference in hamstring MTS, these geometric characteristics only explained 16% of the variance in MTS. In combination, these findings indicate that while muscle size undoubtedly influences MTS, a combination of additional factors (e.g., muscle architecture, fiber type distribution, etc.) may explain a substantial portion of MTS variance in addition to that explained by muscle size. Future research is necessary to evaluate the influences that these factors have on MTS in efforts to clarify the derivations of the sex difference in hamstring MTS.

Our data suggest that females' hamstrings possess a limited capacity for resisting changes in length associated with joint perturbation compared to that of males. As such, lesser hamstring MTS may translate into diminished dynamic knee joint stability in females, thus contributing to their higher incidence of ACL injury. Though the sex difference in hamstring MTS is integrally linked to muscle size and not to inherent differences in material composition, it is important to note that great care was taken in development of the experimental design to standardize experimental factors which potentially influence between-subject differences in MTS. Namely, the applied load was standardized across subjects at 10% body weight. Our previous work indicates that this load magnitude requires ~30% maximal contractile effort (i.e., EMG amplitude) to support the shank in the testing position, and that this level of activity does not differ across sex (Blackburn et al., 2004). Additionally, the magnitude of the perturbation used to induce oscillatory knee flexion/extension was similar across sex, as the perturbation force was similar in both groups. The similarity in these experimental factors indicates that in response to the same relative loading conditions, the male hamstrings have an enhanced ability to resist changes in muscle length. As such, males potentially have a more efficient/effective dynamic protective mechanism for the knee joint, a factor which may influence ACL injury risk.

While we are unaware of previous literature regarding the direct influence of hamstring MTS on knee joint stability or ACL injury risk, the mechanical definition of stiffness inherently implies that heightened MTS would allow for a more stable joint. Of particular interest is the notion that MTS can be enhanced via training and rehabilitation (Kubo et al., 2001; Burgess et al., 2007), thus it is a modifiable factor of the neuromuscular system which may prove invaluable for ACL injury prevention efforts. Additionally, preliminary data from our laboratory (Norcross et al., 2008) indicate that hamstring MTS plays an important role in defining contractile properties. Specifically, the rate of hamstring isometric force production and time required to attain 50% maximal isometric force were significantly greater in males than in females, and were related to MTS. Assuming that a critical level of contractile force exists for the maintenance of joint stability, these data may suggest that males can achieve this critical force earlier in response

to joint perturbation. Additionally, hamstring stiffness was measured in the current investigation with the hamstrings active under a specified submaximal condition. MTS increases with the level of contraction effort (McNair et al., 1992), and the rate of increase in stiffness per unit of increase in contraction effort is greater in males than in females (Granata et al., 2002). Accordingly, the potential sex discrepancy in joint stability as a function of MTS may be greater at higher levels of contractile effort.

Several limitations to the model incorporated in this investigation merit discussion. First, our estimates of hamstring CSA were anatomical in nature rather than physiologic. As such, our values for hamstring stress, and therefore elastic modulus, may have been overestimated. As mentioned previously, our calculation of hamstring length based on digitized coordinates for the origin and insertion ignores the curvature/wrapping of the musculotendinous unit (Arnold and Delp, 2001), as the biceps femoris tendon wraps infero-laterally around the proximal posterior tibia and fibula. While this limitation likely underestimates the true length of the biceps femoris long head, the magnitude of this difference is likely minimal. An additional limitation to our model is the assumption of equal force contributions from the medial and lateral hamstrings during the MTS assessment. Though previous biomechanical knee models have operated under this assumption (Kellis et al., 2005), the true relative contributions of the lateral and medial hamstrings are unknown. However, Fiebert et al. (1992) reported similar EMG amplitudes between the medial and lateral hamstrings during isometric contractions. In a more recent investigation (Fiebert et al., 2001), these authors reported a significantly greater contribution (based on EMG) from the lateral hamstrings during submaximal isometric contractions, yet both muscle groups contributed approximately 50% during submaximal contraction (56% lateral and 44% medial). This model also does not account for the role that the gastrocnemius plays in knee flexion moment production. However, we demonstrated previously that the lateral gastrocnemius displays minimal EMG activity throughout this task (Blackburn et al., 2004) and, therefore, suggest that its contribution to our stiffness values is likely negligible. The mass of the shank and foot segment for use in the calculation of MTS was estimated from tabled anthropometric data derived from elderly male cadavers (Winter, 1990). As estimates of segmental anthropometric characteristics may differ across populations (Durkin and Dowling, 2003), the use of these data may have lead to errors in segment mass estimations. Lastly, modeling the knee as a single-degree-of-freedom system assumed the absence of axial plane rotation. While we did not verify this assumption via motion capture, the perturbation was isolated to the vertical dimension as much as possible, and visual observation of the resulting oscillatory motion suggested that axial plane motion was negligible. Though these limitations were inherent to the experimental design, all procedures were identical in all subjects, thus it is unlikely that a systematic bias occurred such that the fidelity of the sex comparisons was compromised.

5. Conclusions

The results of this investigation suggest that the structural properties of the hamstrings differ across sex. Females demonstrated lesser hamstring stiffness compared to males in response to standardized loading conditions, indicating a compromised ability to resist changes in length associated with joint perturbation. However, the difference in hamstring stiffness was at least partially attributable to differences in muscle size, as the material properties of the hamstrings did not differ across sex. Future research is necessary to determine the clinical/physiological implications of hamstring MTS for knee joint stability and ACL injury risk.

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