Gender Differences in Both Active and Passive Parts of the Plantar Flexors Series Elastic Component Stiffness and Geometrical Parameters of the Muscle–Tendon Complex

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ABSTRACT: Men are reportedly at higher risk of plantar flexor muscle injury and Achilles tendon ruptures than women. Biomechanical parameters are thought to play a role in the higher frequency of injury to males. One parameter is the stiffness of tissues; a stiff tissue cannot absorb sufficient energy with loading, and subsequently may be more likely to be injured. Thus, our purpose was to investigate the gender difference in the geometrical parameters of plantar flexor's muscle–tendon complex and the stiffness of both active and passive parts of the series elastic component (S_{SEC1} and S_{SEC2}, respectively). Using the alpha method on data obtained from quick stretches to the plantar flexors performed during isometric contractions, S_{SEC1} and S_{SEC2} were assessed. Plantar flexor muscles and Achilles tendon cross-sectional areas (CSA_{TS} and CSA_{AT}, respectively) were determined in young healthy men ($n = 49$) and women ($n = 31$). The findings showed that S_{SEC2} was higher in men ($p < 0.001$), but this difference was not apparent when S_{SEC2} was normalized to CSA_{AT} ($p > 0.05$). In contrast, S_{SEC1} was lower in men ($p < 0.001$) and remained so after normalization to CSA_{TS}. Higher joint stiffness observed in men was notably influenced by lever arm length. Thus, the results of this study have implications for performance and injury. © 2011 Orthopaedic Research Society. Published by Wiley Periodicals, Inc. J Orthop Res

Keywords: cross-sectional area; stiffness; men; women; injury risk

The incidence of muscle and tendon injury of the plantar flexor muscles is greater in males compared to females. The ratio of Achilles tendon rupture in men to women varies from 2:1 to 12:1.[1] Among the factors that might be responsible, biomechanical parameters, particularly the stiffness properties of the muscle and tendon, are of interest.[2–6] Stiffness refers to the degree of resistance offered by tissues in response to lengthening. Interest in stiffness has arisen from observations that many muscle and tendon injuries occur when the joint is in the mid range and not in an overextended position.[7] The etiology of tendon and muscle strains is multifactorial.[8] Among the factors implicated in these injuries is a loss of extensibility in the soft tissues.[9] When tissues are too stiff, they likely cannot absorb sufficient energy with loading, and subsequently are more likely to be injured.[10,11]

In optimal performance in gait activities, muscle–tendon complex (MTC) stiffness is important,[3,12] as it directly affects elastic energy storage-recoil processes and muscular tension transmission.[13] A high MTC stiffness could be beneficial, for instance, in athletics where rapid rates of force development are required. Furthermore, specific training techniques such as plyometrics increase series elastic component stiffness[14] leading to increased jump performance.

Gender differences in stiffness have focused on tendon, reflecting the technical difficulties with measuring muscle in vivo. Most studies utilized ultrasonography in vivo.[15–20] However, the findings are equivocal. Some studies[16,17,19–21] showed that men have higher tendon stiffness than women, but others[15,18] do not. But these studies were not performed in the same tendon, and subjects differed in age, sport activity levels, and morphological features.

Recent works by Fouré et al.[14,22] allowed the specific stiffness of both active and passive parts of the series elastic component (SEC) to be calculated simultaneously in vivo. Their technique is based upon the Hill and Morgan muscle models[23–26] but has been adapted for in vivo experiments, allowing the stiffness of the passive (mainly tendons–aponeuroses structures) and active elements (contractile elements) to be calculated during the same action. This allows a better understanding of the mechanical behavior and interactions between muscle and tendon structures. Previous methods using in vivo (sinusoidal perturbation and quick release techniques) only determined global musculo-articular or musculo-tendinous stiffness. Thus, Fouré et al.'s work provides the opportunity to explore gender differences in the MTC stiffness more precisely.

Muscle and tendon stiffness are influenced by geometry (cross-sectional area, CSA) and intrinsic tissue properties that can be appreciated indirectly through measurement of Young's modulus. Patellar and Achilles tendon CSAs are greater in men,[17,19,20] but gender differences in Young's modulus have not been consistently reported.[17–20] Geometrical differences in muscle architecture[27] and CSA have been observed.[28] Males have a larger muscle physiological CSA (higher muscle volume, fascicule length, and fascicule angles),[28–30] and furthermore the CSA of type II fibers are larger in men.

Previous studies that assessed tendon mechanical properties considered moment arm length and tendon

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CSA.[17–20] Other studies used muscle CSA to normalize the musculoskeletal complex stiffness of the ankle.[31,32] However, no study has assessed gender difference considering the specific stiffness of the muscle or the active part of the SEC in regard to the muscle's geometry. Therefore, our aim was to explore the source of gender difference in MTC stiffness of the plantar flexors considering the influence of: specific stiffness of active and passive parts of the SEC of the plantar flexors; geometrical parameters and intrinsic mechanical properties of plantar flexors muscle and tendon; and the plantar flexors moment arm length.

MATERIALS AND METHODS

Subjects

Forty-nine men (21.8 ± 3.5 years, 177.7 ± 6.2 cm, 70.0 ± 7.7 kg) and 31 women (22.8 ± 4.2 years, 166.0 ± 5.5 cm, 56.7 ± 5.8 kg) volunteered, but not all subjects were involved in all the parts of the protocol. Fifteen women and six men's muscle CSA were not assessed; thus, their data were not included for analysis of parameters involving muscles CSAs. The number of subjects involved in each measurement is specified in Tables 1–3. Subjects were fully informed about the study and provided signed consent. This study was conducted according to the Helsinki Statement and was approved by the local ethics committee.

Experimental Design

Subjects were tested in a single session to assess: Achilles tendon CSA and length; triceps surae muscle CSA; global angular joint stiffness; linear MTC stiffness; and both active and passive parts of the SEC stiffness.

Achilles Tendon Geometry

Measurements of Achilles tendon cross-sectional area (CSA_{AT}) were obtained from ultrasonographic imaging (Philips HD3, Philips Medical Systems, Andover, MA) with an electronic linear array probe (7.5-MHz wave frequency; L9-5, Philips) as described by Fouré et al.[3] Subjects were in a prone position on an examination couch with legs fully extended and the ankle at 90°. CSA measurements were performed in the transverse plane with the transducer perpendicular to the tendon at the level of the medial malleolus.[33] Achilles tendon length (L_{AT}) was defined as the distance between the most distal portion of the gastrocnemius medialis muscle[34] and the tendon insertion at the calcaneum, measured via ultrasonography.

Triceps Surae Muscles Geometry

Measurement of the plantar flexors muscles CSA was performed using the same ultrasonographic device and the probe described by Fouré et al.[14] Subjects stood on one leg with the ankle and knee of the other leg flexed at 90°, and fully immersed in a water-filled container. Using a specific guide, 12 images were taken transversally (4 images on each side of the lower leg) at 50%, 60%, and 70% of lower leg length, as defined as the distance between the center of the lateral malleolus and the popliteal crease. The triceps surae cross-sectional area (CSA_{TS}) was averaged across these three measurements; at each point of the lower leg length, all images were assembled to obtain a full lower leg cross-sectional area image using PowerPoint (Microsoft Corp., Redmond, WA). CSA_{TS} was determined from these images using open source digital measurement software (Image J, NIH, Bethesda, MD).

Global Angular Joint Stiffness and Linear MTC Stiffness

As described previously,[22] a Biodex system 3 research® (Biodex Medical, Shirley, NY) isokinetic dynamometer was used to measure the external torque, ankle joint angle, and ankle joint angular velocity. Subjects were seated, legs fully extended with thighs, hip, and shoulders secured by adjustable lap belts. The right ankle was securely strapped to a footplate connected to the dynamometer's lever arm. The input axis of the dynamometer was adjusted to the rotational axis of the ankle. Ankle angle was fixed at 75° (in plantar flexion, the foot perpendicular to the tibia = 90°), and the hip was flexed 40°. These signals were sampled at 1,000 Hz and stored on a hard disc for further analysis.

The protocol to assess the ankle joint stiffness included: a warm up consisting of 10 submaximal isometric plantar flexion contractions; 2 maximal voluntary contractions under isometric conditions in plantar flexion performed at 75° with 2 min rest after each trial (the maximal isometric torque (MVC) across trials was utilized in the analysis); and a familiarization to the short-range stiffness experiment. Subjects had to sustain two submaximal torques at levels of 40% and 80% of the MVC. The torque level was maintained by the subject through video feedback and presentation of the level to be attained. When the level was constant, a passive stretch of 20° (from 75° to 95°) was applied at a high velocity (~250°/s) into dorsiflexion. Thereafter, the change in torque and angle (~7°) during the first 60 ms of stretch was considered for further analysis. The short range stiffness experiments (14 trials) were performed at 7 levels of submaximal torque in a random order (2 trials at each 10% of MVC from 30% to 90% of MVC) with 2 min of rest between trials.

Table 1. Maximal Voluntary Contraction in Plantar Flexion (MVC), Plantar Flexors Moment Arm Length (MA), Achilles Tendon Cross-Sectional Area (CSA_{AT}), Length (L_{AT}) and Cross-Sectional Area of the Triceps Surae Muscles (CSA_{TS})

	Men	Women	N (Men/Women)
MVC (N m)	129 ± 19	97 ± 18***	49/31
MA (mm)	51.6 ± 2.3	47.5 ± 2.4***	49/31
CSA_{AT} (mm^2)	58.6 ± 11.7	47.3 ± 10.2***	49/31
L_{AT} (mm)	194 ± 22	171 ± 21***	49/31
CSA_{TS} (mm^2)	3,805 ± 580	3,312 ± 492**	43/16

N, number of subjects. Results are presented as mean ± standard deviation. Significant difference between men and women: $^{**}p < 0.01$, $^{***}p < 0.001$.

Table 2. Global Angular Joint Stiffness and Linear MTC Stiffness (S_{ANG} and S_{LIN}, Respectively) Determined at 30 and 90% of Maximal Voluntary Contraction and for an Absolute Level of Contraction (50 Nm and 1,200 N, Respectively)

		Men	Women	N (Men/Women)
S_{ANG} (N m $^{-1}$)	30% MVC	3.71 ± 0.48	3.01 ± 0.45***	49/31
	90% MVC	6.27 ± 0.83	4.93 ± 0.77***	49/31
	50 N m	4.30 ± 0.27	3.95 ± 0.36***	49/31
S_{LIN} (N mm^{-1})	30% MVC	46.8 ± 5.7	44.8 ± 7.9	49/31
	90% MVC	82.4 ± 9.6	76.2 ± 12.0**	49/31
	1,200 N	61.7 ± 4.3	63.9 ± 69*	49/31

N, number of subjects. Results are presented as mean ± standard deviation. Significant difference between men and women: $^{*}p < 0.05$, $^{**}p < 0.01$, $^{***}p < 0.001$.

Data processing was described previously.[14,22] In brief, the measured torque was corrected for inertia and the dynamometer's weight arm to obtain the external torque at the ankle. The external torque and the ankle angle were recorded when the joint commenced motion (when velocity > 0) to 60 ms thereafter. The global angular joint stiffness (S_{ANG}) was determined as the ratio between torque and angle changes. S_{ANG} was determined at 50 Nm, 30% and 90% of MVC.

To account for the moment arm length difference between genders, the plantar flexors MTC length and moment arm length (MA) were estimated from ankle and knee joint angles and the limb length of each subject.[35] External force (F) was estimated from external torque using:

$$F = \frac{T}{\mathrm{MA}} \quad (1)$$

The linear MTC stiffness was determined as the ratio between force and MTC length changes.

Active and Passive Parts of the Series Elastic Component Stiffness

The joint compliance (inverse of the linear MTC stiffness) was considered as the compliance of two springs in series, one representing compliance of the active and the other the passive part of the series elastic components. The active compliance was regarded as inversely proportional to the force (F),[23] and the passive compliance (C_{SEC2}) was assumed constant over the range of torque considered (from 30% to 90% of MVC).[14] Thus, the relationship between ankle joint compliance and force is:

$$\alpha = C \times F = \alpha_0 + C_{SEC2} \times F \quad (2)$$

where α is the product F and ankle MTC compliance (C), and α_0 is the elastic extension with the force dependent component of the series elastic component (SEC).

Thereafter, a linear regression was applied to establish the relationship between α and F. α_0 and C_{SEC2} were then the Y-intercept and slope, respectively. α_0 and C_{SEC2} were used to calculate the modeled linear MTC stiffness (S_{LIN}):

$$S_{LIN} = \frac{F}{F \times C_{SEC2} + \alpha_0} \quad (3)$$

The joint stiffness–torque relationship was assessed for each subject. A stiffness index of the active part of the SEC and stiffness of the passive part of the SEC (S_{SEC2}) were also calculated as the inverse of α_0 and C_{SEC2}, respectively. S_{LIN} was also determined at 30% and 90% of MVC and for an absolute force of 1,200 N, chosen because it was a common force that all subjects attained and is typical of loads experienced during daily activities.[36,37] The stiffness of the active part of the SEC (S_{SEC1}) was also determined at 1,200 N and MVC.

Reliability

The reliability of the stiffness properties (S_{ANG}, SI_{SEC1} and S_{SEC2}) and geometrical parameters (L_{AT}, CSA_{AT}, and triceps surae muscle CSA) was assessed previously.[14,22] Two testing sessions completed at the same time of the day with 2 days rest in between were performed to determine reliability. Intraclass correlation coefficients >0.88 and coefficients of variation <6.0% were observed.

Table 3. Mean Values of the Passive Part of the Series Elastic Component Stiffness (S_{SEC2}), Normalized with Achilles Tendon Cross-Sectional Area (CSA_{AT}) and the Active Part of the Series Elastic Component Stiffness (S_{SEC1}) Determined at 1,200 N and for Maximal Voluntary Contraction ($S_{SEC1\ 1,200N}$ and $S_{SEC1\ max}$, Respectively), both Normalized with Triceps Surae Muscle Cross-Sectional Area (CSA_{TS})

	Men	Women	N (Men/Women)
S_{SEC2} (N mm^{-1})	135.2 ±23.7	121.1 ± 25.5**	49/31
S_{SEC2}/CSA_{AT} (N mm^{-3})	2.41 ± 0.73	2.65 ± 0.73	49/31
$S_{SEC1\ 1,200\ N}$ (N mm^{-1})	118.0 ± 18.4	149.6 ± 42.2***	49/31
$S_{SEC1\ max}$ (N mm^{-1})	244.2 ± 43.6	249.8 ± 67.4	49/31
$S_{SEC1\ 1,200\ N}/CSA_{TS}$ (N mm^{-3})	0.0322 ±0.0076	0.0429 ± 0.0164***	43/16
$S_{SEC1\ max}/CSA_{TS}$ (N mm^{-3})	0.0662 ± 0.0141	0.0723 ± 0.0217	43/16

N, number of subjects. Results are presented as mean ± standard deviation. Significant difference between men and women: $^{**}p < 0.01$, $^{***}p < 0.001$.

Statistics

After checking the data distributions using a Shapiro–Wilk test, parametric tests were performed using Statistica® software (Statsoft, Inc., Tulsa, OK). Student's *t*-tests were performed to assess the significance of gender on all variables with level of significance set at $p < 0.05$.

RESULTS

MVC was significantly greater in men ($p < 0.001$; Table 1). Women had a shorter, thinner Achilles tendon ($p < 0.001$) compared with men. Men had a higher CSA_{TS} and moment arm ($p < 0.01$).

Global angular joint stiffness was significantly higher in men ($p < 0.001$) irrespective of absolute or relative levels of torque (Table 2). The relationships between global angular joint stiffness and torque for men and women are shown in Figure 1A. However, when the moment arm was considered, no significant difference was found between gender for linear MTC stiffness determined at 30% MVC ($p > 0.05$); a higher linear MTC stiffness determined at 90% MVC was found in men ($p < 0.05$) and a significant higher linear stiffness was found for women ($p < 0.05$) when determined at 1,200 N (Table 2). The relationship between linear MTC stiffness and force for men and women is shown in the Figure 1B.

S_{SEC2} was significantly higher in men ($p < 0.001$; Table 3). However, this difference was not apparent when S_{SEC2} was normalized to CSA_{AT} ($p > 0.05$; Table 3). In addition, S_{SEC1} determined for 1,200 N was significantly higher in women even when this parameter was normalized to CSA_{TS} ($p < 0.01$; Table 3). The relationships between absolute and normalized S_{SEC1} and force are shown in Figure 2.

DISCUSSION

We aimed to determine whether gender differences were apparent in stiffness and geometrical parameters of the plantar flexors MTC, with a view that these biomechanical factors might provide clues as to why men have more injuries to this muscle group compared to females. Our results showed a higher stiffness of the passive part of the SEC in men. In contrast, women had higher stiffness of the active part of the SEC even when S_{SEC1} was normalized to CSA_{TS}.

Men had higher global angular joint stiffness than women as previously shown in studies using sinusoidal perturbations.[31,38–40] Many authors[39–42] focused upon differences in mechanical stress applied during work and recreational activities and/or the effects of hormones on tissues synthesis and degradation mechanisms. However, the influence of lever arm was not considered previously in MTC stiffness assessment. This length played a notable role in this difference, as

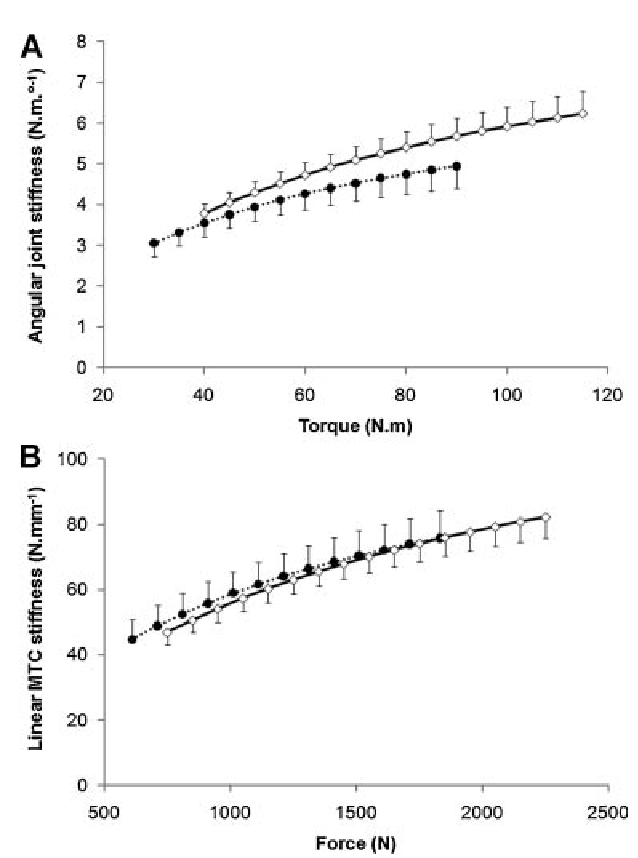

Figure 1. Mean global angular joint stiffness–torque (A) and linear MTC stiffness–force (B) relationships obtained for men (—◇—)andwomen(··●··).

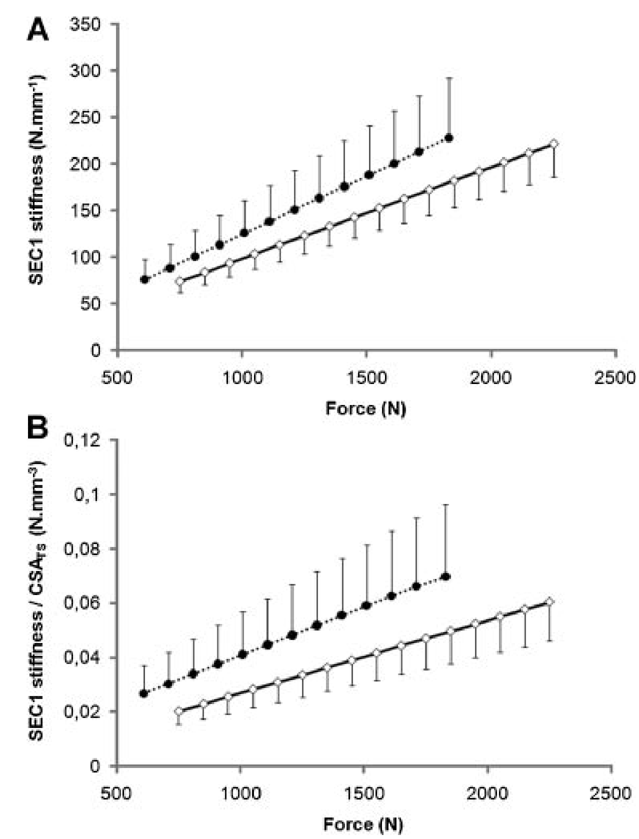

Figure 2. Mean relationships between active part of the series elastic component stiffness (S_{SEC1}) and force (A) and S_{SEC1} normalized with *triceps surae* muscles cross-sectional area (CSA_{TS}) and force (B) relationships obtained for men (—◇—)andwomen(··●··).

indicated in Figure 1 which shows that linear stiffness is similar between genders.

In our study, normalization of the S_{SEC2} to the Achilles tendon CSA was considered an approximate indicator of the Young's modulus of the tendinous tissues. Studies in the literature concerning gender differences in modulus are conflicting. For instance, Kubo et al.[17] and Burgess et al.[21] found a significant difference between men and women in Achilles tendon modulus, whereas Onambele et al.,[19] Westh et al.,[20] and O'Brien et al.[18] found no significant difference in the modulus of patellar tendon. Our result concerning the S_{SEC2} is different from results obtained in Achilles tendon stiffness. The Achilles tendon per se is only part of the passive component of the SEC. Aponeuroses have different mechanical properties than tendon.[43] The relative gender difference determined in our study in S_{SEC2} ($\sim$10%) is lower than those of previous studies performed on Achilles tendon ($\sim$50%).[17,21] Nevertheless, according to our results, geometrical parameters are a primary factor influencing gender differences in the passive part of the SEC stiffness rather than intrinsic structural properties of the tendon/aponeuroses tissues. The increased tendon CSA in men is likely related to increased force generation capacity in the muscle fibers, and the Achilles tendon subsequently adapts to meet the stresses associated with that capacity.

In our study, S_{SEC1} was higher in women at 1,200 N of load ($p < 0.001$). A significant difference was also found in triceps surae muscle CSA ($p < 0.05$), in accordance with previous studies[32,44] performed on tibialis anterior and biceps femoris muscles. However when the S_{SEC1} data were normalized to the CSA, greater stiffness was still apparent in women. Such a difference might be related to the distribution of fast and slow twitch fibers in the plantar flexors. Since fast fibers are more compliant than slow twitch fibers, women might have a greater percentage area of the latter fibers[45–47] and a greater percentage area of type I fibers.[46,48,49] Such findings have been linked to differences in basal mRNA content.[50]

Due to a higher angular stiffness, the musculo-articular system of the plantar flexors of men is less able to absorb energy in men. However, higher joint stiffness in men is mainly explained by geometrical factors and moment arm length is of primary importance (Fig. 1). The higher lever arm for men indicates that the strain on the tissues when stretched is higher for a given change in joint angle, partly explaining the increased risk of injury for men. Even so, the level of stiffness that might be regarded as too great and hence indicative of causing injury. This requires further examination. Although women have a higher ankle flexibility,[51,52] we found a higher stiffness in SEC_1 for women. Therefore, while our findings do not provide a clear answer as to whether gender differences in intrinsic stiffness of tissues can be implicated as a risk factor for soft tissue injury, geometrical factors such as the lever arm may play an important role.

In summary, we investigated differences in mechanical and geometrical properties of each part of the plantar flexors SEC structures. A higher S_{SEC2}, was observed in men, but this was not apparent when the data were normalized to CSA_{TS}. S_{SEC1} was significantly greater in women, and this difference remained when the data were normalized to CSA. These gender differences could affect MTC behavior during function (tension transmission and the storage-recoil process of elastic energy); however, their relationship to soft tissue injury is less clear. Geometric differences such as lever arm length may be important.

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