Gender-specific in vivo measurement of the structural and mechanical properties of the human patellar tendon.

ONAMBÉLÉ, G.N.L., BURGESS, K. and PEARSON, S.J.

2007

Gender-Specific In Vivo Measurement of the Structural and Mechanical Properties of the Human Patellar Tendon

Gladys N.L. Onambélé,[1] Katherine Burgess,[2] Stephen J. Pearson[2]

[1]Department of Exercise & Sport Sciences, Manchester Metropolitan University, Hassall Road, Alsager ST7 2HL, United Kingdom

[2]Centre for Rehabilitation and Human Performance Research, Directorate of Sport, University of Salford, Manchester M6 6PU, United Kingdom

ABSTRACT: Human patellar tendon stress (s), strain (e), stiffness (*K*), and tensile or Young's modulus (*E*), are determined in vivo through voluntary isometric contractions monitored with B-mode ultrasonography. The limitations in previous studies are: (1) they have generally not accounted for the fact that the distal attachment of the patellar tendon (the tibial tuberosity) also displaces; thus, they have underestimated e (and, hence, injury risk) while overestimating *K*; (2) no gender effect has been studied despite the fact that females are seen to have higher incidences of tendon-related injuries. The current investigation therefore aimed to determine the gender specific values of s, e, *K*, and *E* of the patellar tendon while also accounting for distal displacement of the patellar tendon. Healthy young males (aged 23.1 ± 1.3 years, $n = 10$) and females (aged 21.3 ± 0.9 years, $n = 10$) were tested. The maximal e of the young males was ~5–10% higher than that reported in earlier literature. Average female versus male values for e, s, *K*, and *E*, taken at the same force level as the males for comparison purposes, were respectively 10.6 ± 1.0 versus 9.0 ± 1.0%, 36.9 ± 1.4 versus 28.9 ± 0.9 MPa, 1053 ± 108 versus 1652 ± 216 N · mm^{-1}, and 0.61 ± 0.08 versus 0.68 ± 0.10 GPa ($p < 0.05$). There are gender differences in tendon structural and mechanical properties. The current methodology may be useful in a clinical context where early prediction of injury risk and/or monitoring of reconstructed tendon needs to be an accurate, objective, and reliable method if optimal functionality is to be achieved.

Keywords: gender; human patellar tendon; mechanical properties

INTRODUCTION

The tendon is a viscoelastic tissue that deforms in a nonlinear manner in response to the applied loading characteristics. The magnitude of this deformation is affected by the intrinsic stiffness of, and total loading forces applied to, the tendon.[1–3] Therefore, it can easily be understood that explosive sporting events or exercises where tendons are subjected to high loads provide the potential for increased risk of injury due to the subsequent tendon deformation.

Stiffness is a property of the tendon associated with the ability of the tendon to transfer forces rapidly and effectively; a stiffer tendon being able to transfer the muscle forces to the bone more rapidly than a less stiff (i.e., a compliant) tendon. Strain, a measure of the ratio of the lengthening of a tendon upon force application, to the resting length of the same unit, is often used as a measure of injury risk. Indeed, a tendon that continues to deform under load may reach its elastic limit and rupture.[4] This is because tendons have a characteristic elastic range when considering the change in length due to the application of force, whereby if the tendon is strained (i.e., elongated) beyond this region there is an increased risk of rupture. Typically, strains of 23–30%[4] and as low as ~14%,[5] have been reported to cause rupture in patellar tendons in vitro. Most of the current data on tendinous structures tends to be from cadaveric[6] or animal studies.[7,8] Data from these studies do not represent ideal references to living human tissues not least because the in vivo interaction between the muscle and the tendon are not replicated in this type of work. For instance, previous in vitro work in human patellar tendon describe structural properties (e.g., ~15% maximal strain[5]) that are at odds with previous in vivo data (e.g., ~6% maximal strain[9,10]), thereby highlighting the nontransferable nature of the results of in vitro studies to the in vivo events.

Previous studies have utilized ultrasound to determine the in vivo patellar tendon mechanical and structural properties in males by recording the deformation or excursion of the tendon via real-time monitoring of the displacement of the apex of the patellar, under ramped isometric loading conditions.[2,3,9–12] In vivo data, however, is scarce regarding the possibility that both proximal and distal patellar attachments may displace upon voluntary force application. This is indeed a problem if accurate values of strain, and hence, structural tendon properties, and the subsequent estimation of injury risk, is to be accurate.

Furthermore, no previous data exists for in vivo measurement of female patellar tendon structural and mechanical properties, despite the fact that it has been shown that in the gastrocnemius tendon females differ significantly with respect to their viscoelastic properties compared to males in that they exhibit greater tendon elongation, greater strain, lower stiffness, and lower hysteresis.[13] Similarly, indications are that female athletes generally exhibit greater knee laxity,[14] and higher incidence of anterior cruciate ligament injuries than males (for a review see ref. [15]). Hormones, particularly estrogens, have been suggested by some studies to be cause for the disproportionate tendon/ligament/joint laxity in females (for a review see ref. 16).

In addition, previous work has shown the patellar tendon to be a highly adaptable unit whereby stiffness increases with muscle loading[10] and decreases with unloading.[17] Thus, as activity levels are likely to be different between athletes and recreationally active populations, "normative" data should be made available to allow the clinician to have a reference point for comparison in the clinical context.

Therefore, the aims of the present study were to accurately assess the in vivo structural and mechanical tendon properties in a young cohort, using ultrasound as a noninvasive objective method to (1) determine the degree to which the standard method of assessing only proximal displacement underestimates the measured tendon strain compared to when distal displacement is accounted for. (2) Investigate any gender differences on the measured patellar tendon structural and mechanical properties. Our hypothesis being that measured tendon strain would be greater when total tendon elongation is accounted for, thus exposing the fact that previous works have under-estimated strain (ε), and overestimated stiffness (K). We also hypothesized that, partly owing to tendon anatomical dichotomy and partly owing to differences in intrinsic tendon mechanical properties, females would show a greater tendon excursion for any given force in comparison to males.

MATERIALS AND METHODS

Participants and Experimental Design

Twenty recreationally active participants (people accustomed to exercising up to three times weekly), 10 males aged (mean ± SEM) 23.1 ± 1.3 years, height 181.0 ± 2.5 cm, and weight 82.3 ± 4.3 kg, and 10 females 21.3 ± 0.9 years, height 166.6 ± 2.2 cm, and weight 66.9 ± 3.5 kg, volunteered to take part in the study. Owing to the possibility of menstrual cycle phase influence on tendon properties,[16] the female participants only included current oral contraceptive users, and these were not tested during menses so that any gonadal hormones fluctuations were kept to a minimum. Participants had no history of lower limb injury, nor did they have any history of eating disorders, either of which would have prevented them from participating in the study. On the day of testing, participants were assessed at a time that corresponded to approximately 12 h after any exercise, and also tallied with being about 1 h in the postprandial state. The investigation was approved by the local University Institutional Ethics Committee, and all participants gave their written informed consent to participate in the study. The study is in agreement with the declaration of Helsinki of the World Medical Association. Participants visited the laboratory prior to the test session to allow familiarization with the protocols. All measures were preceded by a standardised warmup.

Maximal Patellar Tendon Isometric Force

All measurements of torque were carried out on a Kin Com dynamometer (type 125 AP, Chattanooga, TN). The knee was fixed at 90° flexion (full extension = 0°) and hip at 85° (supine = 0°). The center of rotation of the dynamometer lever arm was aligned with the knee joint center, and straps were fixed across the chest, hip, and thigh of test limb to prevent any extraneous movement. A lever attachment cuff was placed on the lower leg at ~3 cm above the medial malleolus. Participants were instructed to perform ramped isometric knee extensions to maximum over a 4-s period. Tendon force was calculated as previously.[2,3] Briefly, $F_{Isom} = (T_{ob} þ T_{ant}) \div P_{MA}$, where F_{Isom} is the isometric tendon force, T_{ob} is the observed isometric knee extensor torque, T_{ant} is the antagonists (hamstring muscles) cocontraction torque (see section below for the computation of the latter), and P_{MA} is the patellar tendon moment arm (44.7 ± 1.6 mm computed from the average values of previous reports).[18,19]

Estimation of Cocontraction Using Electromyographical (EMG) Activity

A pair of self-adhesive Ag-AgCl electrodes ~15 mm in diameter (Medicotest, Rugmarken, Denmark, type N-10A), was placed on clean, shaved, and previously abraded skin, in a bipolar configuration with a constant between-electrodes distance of ~20 mm, at the distal one-third of the length, in the mid-sagittal plane of the biceps femoris muscle (BF). The reference electrode (Type Q-10A) was placed on the lateral tibial condyle. The raw EMG signal was preamplified (x2000, Neurolog remote AC preamplifier, type NL 822, Digitimer, UK), amplified (x2) (Neurolog isolation amplifier NL 820, Digitimer), bandpass filtered between 500 and 10 Hz (Neurolog, type NL 134 and NL 144, Digitimer), and sampled at 2000 Hz. All EMG and torque signals were displayed in real time in Testpoint software (CEC, MA) via a PC. A series of three maximal flexion contractions were carried out to obtain the EMG at maximal flexion torque. The root mean square (RMS) EMG activity corresponding to the peak torque period was analysed over 50-ms epochs and averaged for a 1-s period during the plateau of peak torque. This has been previously suggested to be acceptable in terms of signal to noise.[20] The EMG of the long head of the BF was measured to ascertain the level of antagonistic muscle cocontraction during the required isometric knee extension performances. Assumptions were that BF is representative of its constituent muscle group,21 and BF EMG relationship with knee flexors torque is linear.[22] BF EMG activity during knee extension was divided by the maximal BF flexor EMG, and the maximal flexor torque was then multiplied by this value to determine cocontraction torque.

Measurement of Tendon Structural and Mechanical Properties

All measurements of the tendon were carried out after a series of three preconditioning trials to ensure reproducibility.[23]

Force–Elongation Relationship

Tendon force–elongation characteristics were determined using a method previously described.[2,3,12,24,25] Briefly, ramped maximal isometric knee extension contractions were carried out at a knee angle of 90° over ~4 s while simultaneously recording generated torque and either (a) patellar proximal (apex of the patellar) or (b) tibia distal (tibial tuberosity) excursions. Patellar tendon imaging was carried out using a B-mode ultrasound unit (AU5, Esaote, Genoa, Italy). The ultrasound probe (7.5 MHz linear array probe, 38 mm wide) was positioned in the sagittal plane over the patellar tendon and the above mentioned anatomical points of interest and tendon excursions were determined at intervals of ~10% of the maximal force. Images were captured in real time at 25 frames per second, directly to a PC (Adobe Premier pro Ver.2) and synchronized (using an electronic signal generator) with the force and EMG records. Image data measurements were processed using digitizing software (Image J, National Institute of Health, Bethesda, MD). Data for each participant consisted of the average of all three trials following the preconditioning contractions.

Tendon Stiffness

Total tendon displacement was computed as the sum of the tibial and patellar displacements. The tendon force–displacement relationships were fitted with second order polynomial functions forced through zero (Fig. 1). Tendon stiffness measures (K in $N \cdot mm^{-1}$), were calculated from the slope of the tangents at 10% force intervals (Fig. 2).

Tendon Tensile Modulus or Young's Modulus

Patellar tendon cross-sectional area (T_{CSA}) and resting length (T_L) were also assessed with the knee joint at 90°. T_{CSA} was measured from the transverse-plane images taken at the average of 0, 25, 50, 75, and 100% of T_L. T_L was measured from sagittal-plane ultrasound images and measured from the inferior pole of the patellar to the superior aspect of the tibial tuberosity. Tensile modulus, hereafter referred to as Young's modulus, was computed as follows: $E = K \times (T_L \div T_{CSA})$.

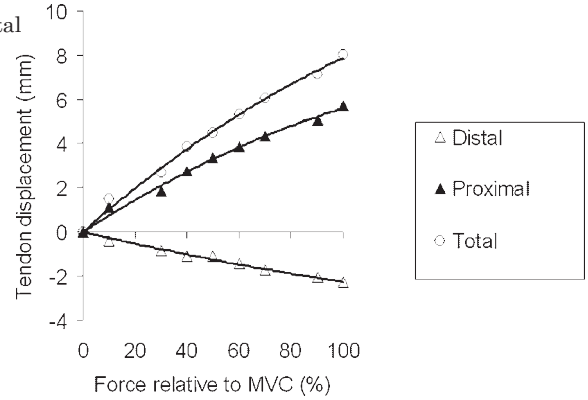
Tendon Strain and Tendon Stress

Tendon strain (%) was calculated as the ratio of tendon excursion to the T_L. Tendon stress (MPa), was calculated by dividing force in the tendon by T_{CSA}.

Statistical Analyses

Tests for differences between the mean were carried out using Student *t-tests* (paired for the methodology comparisons, unpaired for the gender comparisons). Reliability statistics were computed as intraclass correlation coefficients (ICCs) using a one-way random effects model. Significance was set to $p < 0.05$. Effect sizes (r) were calculated to illustrate the importance in practical terms of the detected mean differences. All data are presented as mean ± standard error of the mean (SEM).

Figure 1. Participant sample data for the tibia, patellar apex, and total patellar displacements during a ramped isometric contraction.



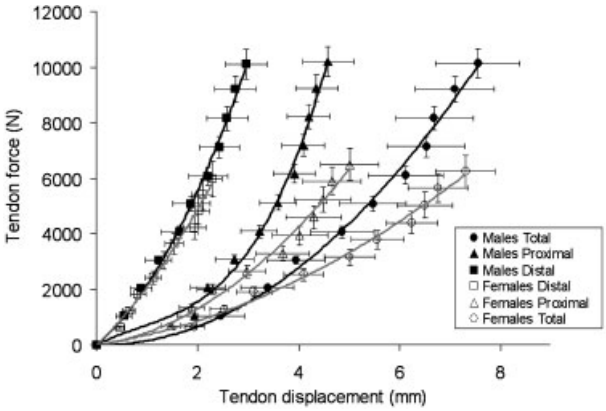


Figure 2. In vivo human patellar tendon force–displacement relationships.

RESULTS

The within-session ICCs were 0.91 for tendon elongation, 0.96 for tendon T_{CSA}, 0.99 for knee extension torque and 0.98 for RMS_EMG.

Females had tendons, which, although similar in length compared with those of the male participants (female 48.9 ± 2.7 vs. males 46.5 ± 2.0 mm, $p > 0.05$), were in fact thinner (female 88.6 ± 5.4 vs. males 112.5 ± 5.9 mm^2, $p < 0.05$, $r = 0.67$). A significant increase in maximal tendon displacement and strain was observed when distal movement was accounted for compared with where only proximal displacements were measured (tendon elongation ±49 ± 1%, $p < 0.0001$, $r = 0.97$, for females and ±66 ± 2%, $p < 0.0001$, $r = 0.93$ for males; strain ± 30 ± 3%, $p < 0.0001$, $r = 0.98$ for females and ±38 ± 2%, $p < 0.0001$, $r = 0.91$ for males). Both maximal total patellar tendon displacement (F, 7.3 ± 0.6 mm; M, 7.5 ± 0.8 mm), and maximal total strain (F, 15.3 ± 1.4%; M 16.5 ± 1.9%, $p > 0.05$) were very similar between females and males. The gender difference in strain is emphasized when it is considered that the males could produce ~62% ($p < 0.0001$, $r = 0.76$) greater MVC than their female counterparts. In fact, maximum total stress was 26% greater for the males (91.3 ± 6.4 MPa) compared to the females (72.3 ± 8.4 MPa).

The average tendon force–elongation relationships were fitted with second order polynomials (Fig. 2). When the displacement of the tibia is taken into account, the force–displacement curve is shifted to the right for both groups and becomes less steep. Both the male and female participants showed significant overestimation of aver- age K values (94 ± 1% for males, and 58 ± 2% for females, $p < 0.0001$, $r = 0.98$) with the method that did not account for the distal tendon displacements. Table 1 also gives the results for patellar tendon stiffness at all levels of force, which suggested a 119% ($p = 0.001$, $r = 0.77$) average gender difference if using the proximal method (i.e., 3759 ± 421 N · mm^{-1} for males and 1716 ± 183 N · mm^{-1} for females, across all forces levels, or a lesser ~80% ($p < 0.001$, $r = 0.68$) gender difference if using the total method (i.e., 1940 ± 221 N · mm^{-1} for males and 1080 ± 110 N · mm^{-1} for females). To account for the differences in tendon dimensions between the groups Young's modulus was calculated (see Table 1), which normalises tendon stiffness values for tendon length and CSA.

Table 1. Effect of Methodology (Proximal Only vs. Proximal þ Distal Displacement Monitoring) on the Computation of In Vivo Structural and Mechanical Tendon Properties

		Displacement (mm)		Stiffness (N · mm^{-1})		Young's Modulus (GPa)	
Gender	%MVC	Proximal Only	Total	Proximal Only	Total	Proximal Only	Total
Males	10	1.8 ± 0.25	2.4 ± 0.39	1590	823	0.66	0.34
	30	2.6 ± 0.31	3.9 ± 0.50	2659	1368	1.10	0.57
	40	3.1 ± 0.29	4.8 ± 0.55	3317	1704	1.37	0.70
	50	3.5 ± 0.30	5.4 ± 0.59	3815	1927	1.58	0.80
	60	3.8 ± 0.36	6.0 ± 0.68	4240	2165	1.75	0.90
	70	4.0 ± 0.37	6.4 ± 0.72	4487	2315	1.85	0.96
	90	4.3 ± 0.37	7.0 ± 0.72	4825	2524	1.99	1.04
	100	4.5 ± 0.46	7.5 ± 0.78	5138	2692	2.12	1.11
Females	10	1.5 ± 0.19	1.9 ± 0.20	830	564	0.46	0.31
	30	2.3 ± 0.31	3.1 ± 0.36	1182	762	0.65	0.42
	40	3.0 ± 0.37	4.1 ± 0.39	1470	927	0.81	0.51
	50	3.7 ± 0.53	5.0 ± 0.50	1773	1082	0.98	0.60
	60	4.0 ± 0.56	5.6 ± 0.52	1915	1173	1.06	0.65
	70	4.3 ± 0.54	6.2 ± 0.51	2034	1288	1.12	0.71
	90	4.7 ± 0.58	6.8 ± 0.59	2187	1374	1.21	0.76
	100	5.0 ± 0.57	7.3 ± 0.59	2338	1466	1.29	0.81

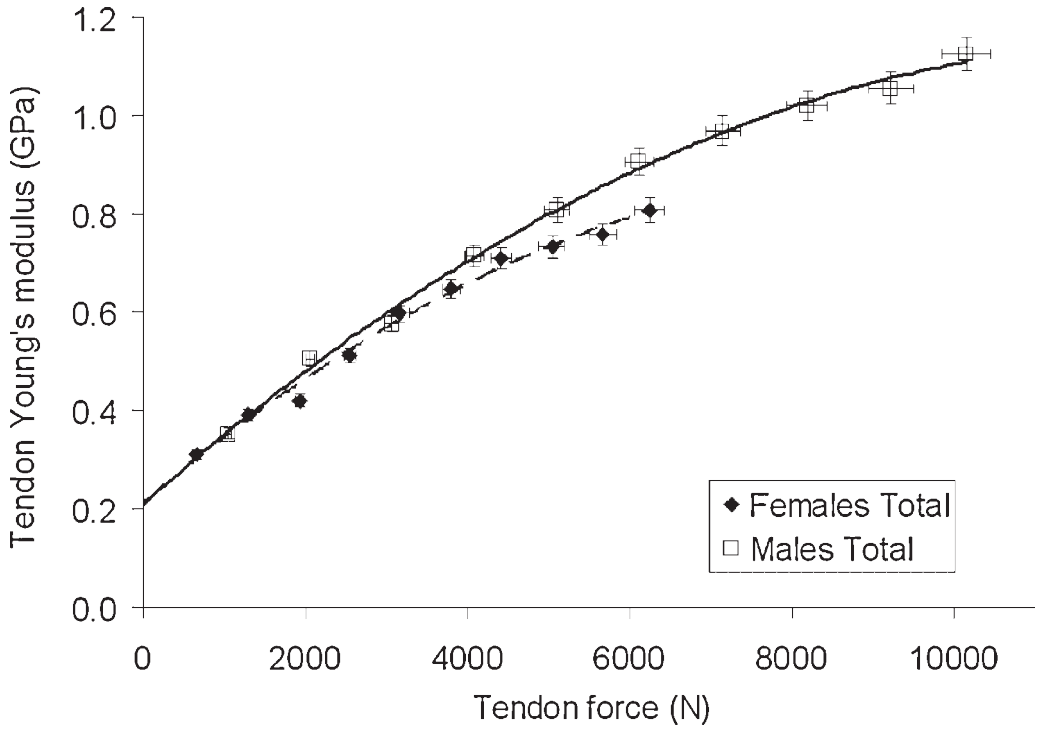


Figure 3. Young's modulus as a function of incrementing level of voluntary tendon forces and the effect of gender. Data are mean ± SEM.

For the total displacement method and at MVC, males were seen to have higher values ($p < 0.05$, $r = 0.66$; see Fig. 3) for both stiffness (2971 ± 389 N · mm^{-1} for males and 1380 ± 187 N · mm^{-1} for females, a 115% gender difference) and Young's modulus (1.2 ± 0.2 for the males vs. 0.8 ± 0.1 GPa for the females, a 53% gender difference, $p > 0.05$).

To account for differences in MVC, a force level of 3230N (corresponding to the maximal MVC of the weakest participant) was used for comparison across genders. Tendon elongation (M, 4.1 ± 0.4 vs. F, 5.1 ± 0.5 cm, a 20% difference, $p > 0.05$), strain (M, 9.0 ± 1.0% vs. F, 10.6 ± 1.3%, a 15% difference, $p > 0.05$), stress (M, 28.9 ± 0.9 vs. F, 36.9 ± 1.4 MPa, a 22% difference, $p < 0.001$, $r = 0.75$) were all found to be lower in the males compared with the females. Tendon stiffness (1652 ± 216 vs. 1053 ± 108 N · mm^{-1}, a 57% difference, $p = 0.023$, $r = 0.51$) and Young's modulus (0.68 ± 0.10 vs. 0.61 ± 0.08 GPa, a 12% difference, $p > 0.05$) at the standardised force level of 3230N were greater in the males compared with the females.

DISCUSSION

The current study aimed to (1) compare patellar tendon elongations during ramped isometric contractions using both proximal and distal tendon excursions, and (2) determine any gender effect. The hypotheses were that measured tendon strain would be greater when total tendon elongation is accounted for, and that, owing in part to anatomical dichotomy and in part to differences in intrinsic material mechanical properties, females would show a greater tendon excursion for any given force in comparison to their male counterparts. Our findings have supported the above hypotheses: (1) the main patellar tendon structural property of interest (i.e., tendon elongation) deter- mined for the males were significantly greater than those in the literature, which had been obtained by observing only the proximal displacement of the patellar tendon. (2) Values for female patellar tendon structural and mechanical proper- ties are significantly different to those for males.

(3) Gender differences persist to some degree even after tendon dimensions are accounted for.

Within the muscle–tendon unit, the series elastic element (i.e., the tendon) is a structure of particular interest in that it is functionally crucial to skeletal muscle contractile characteristics and as such influences all manners of tasks of daily motor performance.[26] An in vivo measure of tendon stiffness (i.e., the computation of the relation between deformations of this structure to the applied forces) is therefore a useful index of the functionality of the tendon. Thus, all things being equal (e.g., tendon dimensions (length and/or cross-sectional area), architecture, and hormonal milieu), a compliant tendon will allow, relatively slow transmission of forces from muscle to bone, thereby exhibiting the tendon as a "viscoelastic damper." A very compliant tendon may adversely affect motor control. Taking the example of postural balance, this deleterious effect of increased compliance is possibly because the rate of the "catch and throw"[27] calf muscle actions required in stance tasks, are delayed. Such an adverse effect of high tendon compliance on indices of postural balance ability has recently been demonstrated in the elderly.[24,25] A direct and severe consequence of poor postural control in the elderly at least, has previously been linked to falls and the subsequent vicious circle of decreased confidence, decreased functional ability and increased morbidity and/or mortality.[28,29]

By reporting here the values for a group of young recreationally active males and females, we have made available comparative data which would be useful as a "gold standard" in monitoring individuals who have, for example, been treated for tendinous injury. It should be pointed out here that clinicians have already attempted to use a measurement of in vivo tendon properties to help determine the risk of rupture of that structure and also to use these properties as indicators of repair after, for example, reconstructive surgery or in the course of the assessment of the tendon injury-risk associated with a sporting activity.[30,31] However, although these techniques also involve the use of ultrasound imaging, they are either not very objective, as they rely on the clinician's "eye" and familiarity with the images to infer the material quality of the tendon,[30] or they involve complicated software that is not yet widely commercially available.[31] The advantage of the method we present here is primarily its objective nature, but also the ease of data interpretation, where normative data is available for comparative purposes. As mentioned earlier in the introduction, previous studies have used a similar methodology, although ignoring the distal tendon displacement.

The only previous study that may compare to ours is that of Hansen et al.,[32] which was the first to show that in vivo, because tibial movement also occurs under voluntary isometric loading conditions, these excursions must be accounted for to accurately determine total tendon strain. These authors quoted a 45 ± 8% underestimation of tendon strain where tibial movement is not monitored, which is somewhat greater than the 38 ± 2% value we report here. Interestingly also, the strain reported by these authors ~6.8 ± 0.7% is lower than that which we found (16.3 ± 1.8%). Aside from tendon dimensional differences (mean T_{CSA}/T_L for Hansen et al.[32] and the current study were respectively ~0.39 and ~0.24), and differences in the maximal force capacities of the males in Hansen et al.[32] versus the current report (~6000N vs 10,141 ± 540 N), it is probable that the disagreements between ours and the value quoted by these authors lays in the fact that the authors used long ramped contractions (10 s) as opposed to our shorter contractions (3–4 s). This is indeed important if we consider that the tendon is a viscoelastic element that is likely to be affected by creep (i.e., the tendency to elongate further if the duration of force application is increased).

This therefore means that, because the current work has used a contraction duration seen in the majority of earlier studies,[2,3,9–12] the value for degree of error in strain reported here, has a greater potential for use in adjusting the strain values reported in said majority of earlier reports. Although the total maximal tendon stress and strain we show here are worryingly close to the values at failures reported using isolated human cadaver tissue (54–65 MPa and 14–15%, respectively[5]), this was to be expected, as said cadaveric tissue was much older (ranging from 29–93 years vs. the current age range of 19–33 years) and was thus expected to be weaker.

This therefore means that, because the current work has used a contraction duration seen in the majority of earlier studies, [2,3,9–12] the value for degree of error in strain reported here, has a greater potential for use in adjusting the strain values reported in said majority of earlier reports. Although the total maximal tendon stress and strain we show here are worryingly close to the values at failures reported using isolated human cadaver tissue (54–65 MPa and 14–15%, respectively[5]), this was to be expected, as said cadaveric tissue was much older (ranging from 29–93 years vs. the current age range of 19–33 years) and was thus expected to be weaker.

In addition to the above, as far as we are aware, the current investigation is also the first to report values of patellar tendon structural and mechanical properties for a group of young females. Fundamentally, we have shown that in our group of recreationally active females, the strain values (15.3 ±1.4%) for the patellar tendon were within the values previously reported as rupture inducing, that is, they are within the previously reported values for ultimate tensile strain at failure(~14 ± 6% in vitro[4]).

Compared with males, females had relatively greater strain values; this is despite the ability of the males to generate significantly higher forces. It is probable that in the males, the greater force generating capacity was offset by the higher tendon cross-sectional area. The higher values of strain observed for the females relative to their male counterparts is in general agreement with work by Kubo et al.,[13] who examined the structural and mechanical properties of the gastrocnemius tendon in males and females and found females to develop greater strain. Our results also suggest that the gender differences may partly be owing to differences in tendon dimensions, because prior to the dimensional adjustments (tendon stiffness), gender difference was 115%, whereas after adjusting for tendon size and cross-sectional area (Young's modulus), the difference between males and females decreased to 53%. In other words, dimensional differences do not explain all of the gender differences in the structural and mechanical properties of the tendon.

Gender differences in the values of Young's modulus could be partly explained by the suggestion that the composition of the tendons is different. It may be that the cross-linking density or arrangement of these tendons is distinct between the two or that the ratio of type I to type III collagen is different. Sensitivity to cellular signals (i.e., Il-1, IL-6, TGF-b), which may increase type I collagen turnover in tendon (for a review see ref. 33), may also be different between the genders, at the same time as the total water content of the tendon could also differ between genders, thereby influencing tendons' intrinsic structural and mechanical properties.

It has also been suggested that circulating hormonal factors such as estrogen and/or progesterone[34] could affect tendon stiffness in that presence rather than absence of any of these hormones has been associated with decreased stiffness (i.e., increased laxity) of ligamentous tissues. In the present study, all females were currently taking the oral contraceptive pill (OCP), ensuring stable circulating levels of at least estrogen and progesterone. However, it is still a possibility that gender differences in absolute circulating levels in these and other hormones could account for some of the observed differences.

Furthermore, the fact that tendon stiffness in the females was seen to be significantly lower than that for males, may indicate a disadvantage in tasks requiring fine motor control, as well as increased risk of falls and fall-related injuries in later life for females relative to males. Indeed, a reduced stiffness would lend itself to greater tendon deformation for equivalent forces, thus increasing the potential for large strain values and subsequent tendon-related injuries. If large forces are experienced, such as in activities where large inertias are being decelerated (e.g., running, jumping, and even walking swiftly down a flight of stairs), an increased risk of injury due to rupture may be present.

In conclusion, this study describes data for the patellar tendon in young, recreationally active males and females, showing females to have significantly greater values of tendon displacement and strain, but lower values of stiffness and Young's modulus. The current data provides useful reference with regard to gender specific tendon structural and mechanical information for those wishing to compare individuals recovering from surgery or at risk of rupture of this structure. Possible limitations to this study include a relatively specific and/small sample to represent males and females. Larger scale studies using this method would allow a more generalised data set to be generated.

ACKNOWLEDGMENTS

The authors are grateful for the financial support from the University of Salford.

REFERENCES

1. Maganaris CN, Paul JP. 1999. In vivo human tendon mechanical properties. J Physiol 521(Pt 1):307–313.
2. Pearson SJ, Onambe´le´ GN. 2005. Acute changes in knee-extensors torque, fiber pennation, and tendon characteristics. Chronobiol Int 22:1013–1027.
3. Pearson SJ, Onambe´le´ GN. 2006. Influence of time of day on tendon compliance and estimations of voluntary activation levels. Muscle Nerve 33:792–800.
4. Haut RC, Powlison AC. 1990. The effects of test environment and cyclic stretching on the failure properties of human patellar tendons. J Orthop Res 8:532–540.
5. Johnson GA, Tramaglini DM, Levine RE, et al. 1994. Tensile and viscoelastic properties of human patellar tendon. J Orthop Res 12:796–803.
6. Huang CY, Wang VM, Pawluk RJ, et al. 2005. Inhomogeneous mechanical behavior of the human supraspinatus tendon under uniaxial loading. J Orthop Res 23:924–930.
7. Dressler MR, Butler DL, Wenstrup R, et al. 2002. A potential mechanism for age-related declines in patellar tendon biomechanics. J Orthop Res 20:1315–1322.
8. Haut RC, Lancaster RL, DeCamp CE. 1992. Mechanical properties of the canine patellar tendon: some correlations with age and the content of collagen. J Biomech 25: 163–173.
9. Reeves ND, Maganaris CN, Narici MV. 2003. Effect of strength training on human patella tendon mechanical properties of older individuals. J Physiol 548:971–981.
10. Reeves ND, Narici MV, Maganaris CN. 2003. Strength training alters the viscoelastic properties of tendons in elderly humans. Muscle Nerve 28:74–81.
11. Kubo K, Kanehisa H, Fukunaga T. 2001. Effects of different duration isometric contractions on tendon elasticity in human quadriceps muscles. J Physiol 536:649–655.
12. Onambélé-Pearson NL, Pearson SJ. 2006. Time-of-day effect on patella tendon stiffness alters vastus lateralis fascicle length but not the quadriceps force-angle relationship. J Biomech [Epub ahead of print].
13. Kubo K, Kanehisa H, Fukunaga T. 2003. Gender differences in the viscoelastic properties of tendon structures. Eur J Appl Physiol 88:520–526.
14. Pollard CD, Braun B, Hamill J. 2006. Influence of gender, estrogen and exercise on anterior knee laxity. Clin Biomech (Bristol, Avon) 21:1060–1066.
15. Dugan SA. 2005. Sports-related knee injuries in female athletes: what gives? Am J Phys Med Rehabil 84:122–130.
16. Zazulak BT, Paterno M, Myer GD, et al. 2006. The effects of the menstrual cycle on anterior knee laxity: a systematic review. Sports Med 36:847–862.
17. Maganaris CN, Reeves ND, Rittweger J, et al. 2006. Adaptive response of human tendon to paralysis. Muscle Nerve 33:85–92.
18. Baltzopoulos V. 1995. A videofluoroscopy method for optical distortion correction and measurement of knee-joint kinematics. Clin Biomech (Bristol, Avon) 10:85–92.
19. Krevolin JL, Pandy MG, Pearce JC. 2004. Moment arm of the patellar tendon in the human knee. J Biomech 37: 785–788.
20. Hermens HJ, Freriks B, Disselhorst-Klug C, et al. 2000. Development of recommendations for SEMG sensors and sensor placement procedures. J Electromyogr Kinesiol 10: 361–374.

21. Carolan B, Cafarelli E. 1992. Adaptations in coactivation after isometric resistance training. J Appl Physiol 73: 911–917.
22. Lippold OC. 1952. The relation between integrated action potentials in a human muscle and its isometric tension. J Physiol 117:492–499.
23. Kubo K, Kawakami Y, Kanehisa H, et al. 2002. Measurement of viscoelastic properties of tendon structures in vivo. Scand J Med Sci Sports 12:3–8.
24. Onambe´le´ GL, Narici MV, Maganaris CN. 2006. Calf muscle–tendon properties and postural balance in old age. J Appl Physiol 100:2048–2056.
25. Onambe´le´ G, Narici M, Rejc E, et al. 2006. Contribution of calf muscle–tendon properties to balance ability in the absence of visual feedback and the effects of ageing. Gait Posture (In press).
26. Zajac FE. 1989. Muscle and tendon: properties, models, scaling, and application to biomechanics and motor control. Crit Rev Biomed Eng 17:359–411.
27. Loram ID, Lakie M. 2002. Human balancing of an inverted pendulum: position control by small, ballistic-like, throw and catch movements. J Physiol 540:1111–1124.
28. Onambélé-Pearson G. 2006. Is there a role for exercise in the prevention of osteoporotic fractures? In: MacAuley D, Best T, editors. Evidence-based sports medicine. Oxford, UK: Blackwell Publishing Ltd.
29. Onambélé-Pearson G, Skelton D. 2006. Exercise and bone health. In: New SA, O'Neill T, Morris R, Skelton DA, editors. Management and prevention of osteoporosis. Oxford, UK: Clinical Publishing Ltd.
30. Morvan G, Busson J, Wybier M, et al. 2001. Ultrasound of the ankle. Eur J Ultrasound 14:73–82.
31. Ofer N, Akselrod S, Nyska M, et al. 2004. Motion-based tendon diagnosis using sequence processing of ultrasound images. J Orthop Res 22:1296–1302.
32. Hansen P, Bojsen-Moller J, Aagaard P, et al. 2006. Mechanical properties of the human patellar tendon, in vivo. Clin Biomech (Bristol, Avon) 21:54–58.
33. Kjaer M. 2004. Role of extracellular matrix in adaptation of tendon and skeletal muscle to mechanical loading. Physiol Rev 84:649–698.
34. Uldbjerg N, Ulmsten U. 1990. The physiology of cervical ripening and cervical dilatation and the effect of abortifacient drugs. Baillieres Clin Obstet Gynaecol 4:263–282.