Rapid Report

In vivo human tendon mechanical properties

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1. The aim of the present study was to measure the mechanical properties of human tibialis anterior (TA) tendon *in vivo*.

2. Measurements were taken in five males at the neutral ankle position and involved: (a) isometric dynamometry upon increasing the voltage of percutaneous electrical stimulation of the TA muscle, (b) real-time ultrasonography for measurements of the TA tendon origin displacement during contraction and tendon cross-sectional area, and (c) magnetic resonance imaging for estimation of the TA tendon length and moment arm.

3. From the measured joint moments and estimated moment arms, the values of tendon force were calculated and divided by cross-sectional area to obtain stress values. The displacements of the TA tendon origin from rest to all contraction intensities were normalized to tendon length to obtain strain values. From the data obtained, the tendon force–displacement and stress–strain relationships were determined and the tendon stiffness and Young's modulus were calculated.

4. Tendon force and stress increased curvilinearly as a function of displacement and strain, respectively. The tendon force and displacement at maximum isometric load were 530 N and 4·1 mm, and the corresponding stress and strain values were 25 MPa and 2·5%, respectively. The tendon stiffness and Young's modulus at maximum isometric load were 161 N mm^{-1} and 1·2 GPa, respectively. These results are in agreement with previous reports on *in vitro* testing of isolated tendons and suggest that under physiological loading the TA tendon operates within the elastic 'toe' region.

Tendons do not behave as inextensible material, but act as biological springs that can stretch elastically storing and releasing energy during locomotion and regulating the muscle mechanical performance (for review see Alexander, 1981; Zajac, 1989). The mechanical properties of tendons have been extensively studied using isolated animal or human material undergoing elongation to failure (e.g. Butler *et al.* 1978; Bennett *et al.* 1986; Lieber *et al.* 1991), but only a few reports exist on tendon mechanical properties under maximal physiological load and most of them refer to animal material testing (Ker *et al.* 1988; Lieber *et al.* 1991). The limited number of such studies on human tendons has indicated that tendon stress, strain and Young's modulus at maximum isometric force are 15–30 MPa, 2% and 1·2 GPa, respectively (Ker *et al.* 1988; Zajac 1989). However, most of these values have been (a) taken from tensile strength testing of preserved or deep-frozen material, which may have altered properties (Smith *et al.* 1996) and be inappropriate when interpreting *in vivo* human function (e.g. Zajac, 1989), and (b) calculated by predicting maximal muscle forces from estimated values of muscle cross-sectional area and maximal stress (Ker *et al.* 1988; Loren & Lieber, 1995). To avoid these problems, assessment of the mechanical properties of intact human tendons has been attempted (e.g. Cook & McDonagh, 1996). However, major assumptions made, e.g. that tendon–aponeurosis stiffness remains constant as a function of force, raise questions about the validity of the results obtained.

Real-time ultrasonography has made possible the recording of collagen-rich tissue within the intact human muscle, offering promise for *in vivo* measurement of changes in muscle–tendon length (Fukunaga *et al.* 1996). Using ultrasonography in the present study, the mechanical properties of the human tibialis anterior (TA) tendon were measured *in vivo*. The methodology presented is applicable to superficially located muscle–tendon units in the human body and allows a direct comparison with results from tests on *in vitro* material.

METHODS

Subjects

Five healthy males, from whom informed consent had previously been obtained, volunteered to participate in this study. None of them had any clinical history of musculoskeletal injury or any orthopaedic abnormality in the lower extremities. Their average (mean ± S.E.M.) age, height and body mass were 22 ± 2 years, 173 ± 3 cm and 75 ± 3 kg, respectively. The study was approved by the local ethics committee.

Experimental design

The mechanical properties of the TA tendon of the right leg were examined at the neutral ankle position where the tibia was at right angles to the sole of the foot. This position was selected because it corresponded to minimal passive forces around the ankle joint (see also Siegler *et al.* 1984). The following steps were taken: (1) measurement of dorsiflexion joint moment; (2) measurement of tendon displacement; (3) calculation of moment arm; (4) calculation of tendon force from (1) and (3); (5) measurement of tendon cross-sectional area and calculation of tendon stress from (4); (6) measurement of tendon length and calculation of tendon strain from (2); (7) calculation of the tendon stiffness and Young's modulus from (2), (4), (5) and (6).

Measurement of dorsiflexion joint moment

Ankle dorsiflexion moments were measured in the prone position on an isokinetic dynamometer (Lido Active, Loredan Biomedical, Davis), having the knee of the tested leg flexed at 90 deg (Fig. 1). Velcro straps were placed over the whole dorsal surface of the foot and the heel, and mounted on the dynamometer footplate. To prevent the retinaculum from stretching upon dorsiflexion contraction (Maganaris *et al.* 1999), inelastic tape was bandaged tightly around the ankle joint. A mechanical stop was placed below the knee to prevent any knee shift. The fixation system used for the foot and knee prevented any observable movement of the lower extremity upon isometric dorsiflexion contraction. The pivot point of the lever arm of the dynamometer was aligned with the rotation axis of the ankle joint and isometric dorsiflexion moments were measured, having compensated for the effect of gravity. To isolate activation of the TA muscle, contraction forces were elicited by means of electrical stimulation. Tetanic contractions of the TA muscle were produced by 1 s of percutaneous stimulation at frequencies of 100 Hz using bipolar wave pulses with a duration of 100 μs. Joint moments were measured at successive stimulating voltages in 25 V steps up to the maximal value. The criteria adopted for the definition of maximal voltage were that either no further increase in joint moment could be achieved with an increasing voltage of 25 V, with plateau values agreeing to within ±5% (Davies *et al.* 1985), or no further voltage could be tolerated by the subject. The electrodes consisted of two aluminium foil pads, 4 cm × 3 cm, covered in tissue which was soaked in water. The electrodes were placed over the motor point area of the muscle, at its proximal and central regions, which had the lowest stimulation threshold resulting in the highest joint moment for a given stimulating voltage. Stimuli were delivered by a custom-built, high-voltage stimulator and controlled by computer software developed in-house (Mr Tom McKee, Manchester Metropolitan University). Surface EMG signals from the nearby soleus and peroneous tertius muscles indicated minimal or no current leakage during stimulation of the TA muscle. Therefore, the joint moments in the study were considered to have been generated by the TA muscle only.

Measurement of tendon displacement

A 7·5 MHz linear-array B-mode ultrasound probe (Esaote Biomedica, Florence; width and depth resolutions, 1 and 0·62 mm, respectively) was placed in the sagittal plane over the distal region of the TA muscle at rest, and the origin of the TA tendon in the musculotendinous junction was identified (Fig. 2). The probe was then oriented along the TA tendon and secured in that position with adhesive tape. With this method of fixation, no shift of the probe could occur upon dorsiflexion contraction. Preliminary experiments in which an echo-absorptive marker was placed between the probe and the skin verified the constancy in the probe position during measurements. The displacement of the TA tendon origin in the transition from rest to all consecutive intensity contractions (i.e. stimulating voltages) was recorded and digitized (Fig. 2). Three measurements were taken 2 min apart and average values were used for further analysis. Displacement data were recorded after the tendon had been pre-conditioned (Bennett *et al.* 1986) by five series of contractions 30 s apart.

Calculation of moment arm

The TA tendon moment arm at the neutral ankle position was estimated using magnetic resonance imaging (MRI) scan morphometrics as described by Maganaris *et al.* (1998*a*, 1999). Sagittal-plane scans of the foot were taken at 15 deg of dorsiflexion, neutral ankle position and 15 deg of plantarflexion (G. E. Signa Advantage 1·5 T/64 MHz, Milwaukee; Fast-GRASS sequence with: repetition time, 15 ms; echo time, 6·7 ms; field of view, 24 cm; number of excitations, 1·0; matrix, 256 pixels × 128 pixels; slice thickness, 5 mm; scanning time, 2 s). In the analysis the ankle joint complex was represented by the tibio-talar joint. In the scan at the neutral ankle position the TA tendon action line was identified and from the scans at 15 deg of dorsiflexion and 15 deg of plantarflexion the centre of rotation in the tibio-talar joint at the

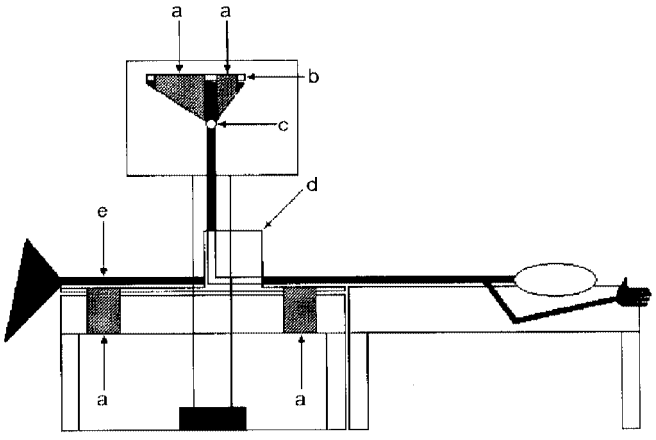

Figure 1. Position of the subject on the dynamometer
The knee is flexed at 90 deg with the foot securely fixed at the neutral ankle position. a, velcro straps; b, dynamometer footplate; c, pivot point of the dynamometer; d, knee mechanical stop; e, contra-lateral limb.

neutral ankle position was calculated using the Reuleaux method (Reuleaux, 1875). The perpendicular distance from the centre of rotation to the tendon line of action was considered as the moment arm. Scans were taken at rest with the ankle joint bandaged as during dorsiflexion joint moment measurements. To assess the effectiveness of the bandage used in maintaining the position of the TA tendon during dorsiflexion loading, additional scans were taken during isometric dorsiflexion produced by maximum voluntary contraction (MVC) in one subject. The scans revealed no visible change in the orientation of the TA tendon in the transition from rest to MVC, suggesting that negligible or no stretch of the retinaculum would have been allowed during electrical stimulation of the TA muscle.

Calculation of tendon force

The force acting along the TA tendon during electrical stimulation was calculated from the moment equilibrium equation around the ankle joint, i.e. by dividing the externally measured joint moment by the calculated moment arm.

Measurement of tendon cross-sectional area and calculation of tendon stress

To reduce tendon forces to stress values, tendon cross-sectional area measurements were obtained using ultrasonography following standard guidelines (Fornage, 1989). With the probe described above, axial-plane scans were taken at five different levels over the length of the tendon: 1 and 4 cm below the level of the TA musculotendinous junction, at the level of the malleoli, and 2 and 4 cm below the level of the malleoli. Cross-sectional areas were digitized and average values from all five sites were used for tendon stress calculation.

Measurement of tendon length and calculation of tendon strain

To calculate strain, measurements of the resting tendon length were taken. Sagittal- and axial-plane MRI scans were taken at the neutral ankle position (Fast Spin Echo sequence with: repetition time, 500 ms; effective echo time, 18 ms; number of excitations, 1; matrix: 256 pixels × 128 pixels, field of view, 48 cm; slice thickness, 10 mm; inter-slice gap, 0 mm; scanning time, 37 s). The TA tendon path and distal point of insertion were visualized in the sagittal-plane scans (Maganaris *et al.* 1998*a*, 1999) and the tendon origin was visualized in the axial-plane scans (Fukunaga *et al.* 1992). The whole tendon length was digitized by following its curved path from origin to insertion.

All MRI and ultrasound scan morphometric measurements in the study were performed three times by the same investigator and mean values were used for further analysis.

Calculation of tendon stiffness and Young's modulus

From the slope of the tendon force–displacement relationship the tendon stiffness was calculated over 50 N tendon force intercepts. The tendon Young's modulus over the respective tendon stress intercepts was then calculated by multiplying the stiffness value obtained by the ratio of tendon length to tendon cross-sectional area.

Measurements of tendon displacement were taken 4 days after a familiarization trial and repeated on two separate occasions 4 days apart (total of 3 experimental trials).

Statistics

Values are presented as means ± S.E.M. Friedman's rank test was used to test differences in the measurements taken (a) as a function of stimulating voltage, and (b) between experimental trials. Statistical difference was set at a level of $P < 0{\cdot}05$.

RESULTS

The maximal stimulating voltage in the study was 150 V. In four of the subjects this was the highest tolerable value while one subject could tolerate 175 V. Application of 175 V in that subject, however, did not result in any joint moment increase compared with 150 V. The dorsiflexion joint moment measured increased from 4·1 ± 1 to 25 ± 3 N m ($P < 0{\cdot}01$) with stimulating voltage increasing from 25 to 150 V

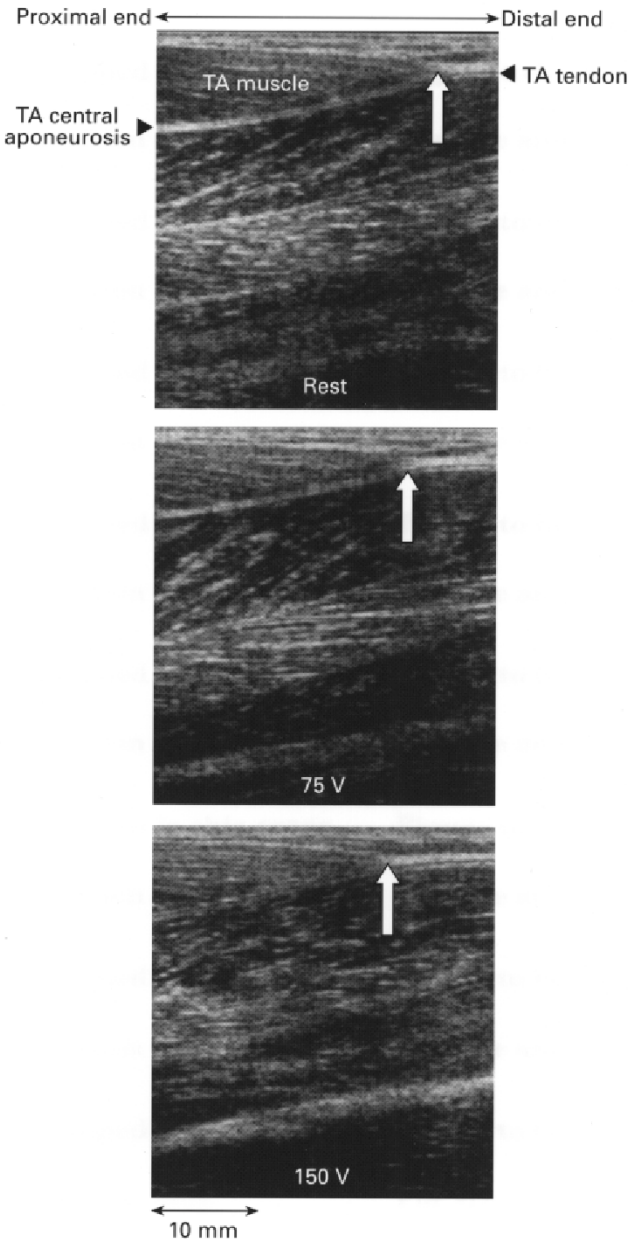

Figure 2. Measurement of tendon displacement

Typical sonographs over the TA musculotendinous junction at rest (top), and electrical stimulation of the TA muscle at 75 V (middle) and 150 V (bottom). The white arrow in each scan points to the TA tendon origin. Notice the displacement of the TA tendon origin in the transition from rest to 75 V contraction and from 75 to 150 V contractions.

(Fig. 3*A*). No difference ($P > 0{\cdot}05$) was found in joint moment measurements between experimental trials.

During electrical stimulation the TA tendon origin shifted proximally (Fig. 2). As the stimulating voltage increased from 25 to 150 V, the displacement of TA tendon origin from its resting position increased from $1{\cdot}1 \pm 0{\cdot}3$ to $4{\cdot}1 \pm 1$ mm ($P < 0{\cdot}01$) (Fig. 3*B*). No difference ($P > 0{\cdot}05$) was found in tendon displacement measurements between experimental trials.

The TA tendon moment arm at the neutral ankle position was 36 ± 4 mm, the tendon length 162 ± 15 mm and the tendon cross-sectional area $20{\cdot}5 \pm 2$ mm^2.

As the stimulating voltage increased from 25 to 150 V, tendon force increased from 71 ± 12 to 530 ± 59 N ($P < 0{\cdot}01$) (Fig. 3*C*). Tendon force increased curvilinearly as a function of tendon displacement, with larger displacements per unit of force at lower forces. The same was the case for the tendon stress–strain relationship. The tendon stress and strain values at 25 V were $3{\cdot}4 \pm 0{\cdot}5$ MPa and $0{\cdot}7 \pm 0{\cdot}1$ %, respectively, and at 150 V they were $25 \pm 2{\cdot}5$ MPa and $2{\cdot}5 \pm 0{\cdot}4$ %, respectively. Representative tendon force–displacement and stress–strain curves from one subject are shown in Fig. 4*A* and *B*, where force and stress values are presented in absolute units. Since tendon force at any given voltage varied across subjects (see above), mean ± S.E.M. values for displacement and strain at given absolute tendon force and stress values, respectively, cannot be obtained directly from our data.

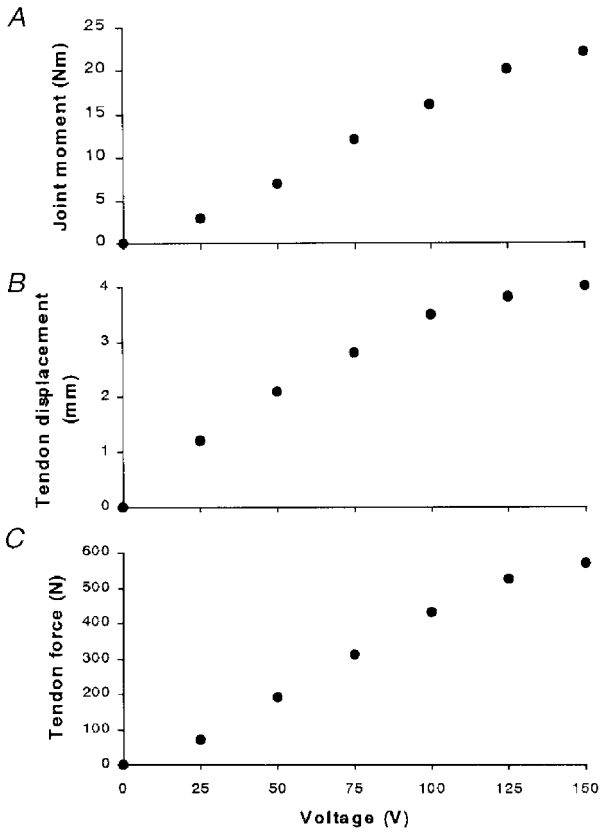

Figure 3. Changes in joint moment, tendon displacement and force as a function of stimulating voltage

A, dorsiflexion joint moment; *B*, TA tendon displacement; *C*, TA tendon force. Data from one subject are presented.

Tendon stiffness increased from 62 ± 12 N mm^{-1} at 0–50 N to 161 ± 26 N mm^{-1} at maximum tendon force (Fig. 4*C*). Young's modulus increased from $0{\cdot}45 \pm 0{\cdot}06$ to $1{\cdot}2 \pm 0{\cdot}15$ GPa with increasing tendon stress (Fig. 4*D*).

Reproducibility of measurements

The TA tendon origin displacement from rest to the 50 V contraction and from rest to the 150 V contraction was recorded and digitized in one subject on 10 occasions after the tendon had been pre-conditioned (see above). The coefficient of variation for the repeat scanning was 6·2% in the former transition and 4·3% in the latter transition. Four different observers recorded and analysed the TA tendon origin displacement in the transition from rest to the 25 V contraction in the same subject and the coefficient of variation for repeat measures was 10·6%. Repeated (15 times) axial-plane scanning of the TA tendon at the level of the malleoli resulted in a coefficient of variation of 3·3% for tendon cross-sectional area. Repeated (3 times) MRI of the foot resulted in a coefficient of variation of 2·9% for TA tendon length. Intra- and inter-observer variations of the TA tendon moment arm measurement have been confirmed to be less than 9% (Maganaris *et al.* 1999).

DISCUSSION

In the present study the mechanical properties of a human tendon were obtained *in vivo* by measuring tendon deformation upon controlled loading as opposed to *in vitro* testing where, inversely, tendon load is measured upon controlled deformation (see also Lieber *et al.* 1991). Before interpreting the results obtained, it is essential to analyse the limitations and assumptions of the methodology followed.

During electrical stimulation the TA muscle–tendon complex was assumed to operate in the sagittal plane only, in which joint moment measurements were taken. The medio-lateral orientation of the TA complex relative to the sagittal plane (Salmons, 1995) would result in an inclined force upon contraction. The ankle joint moment of the TA force vector perpendicular to the sagittal plane cannot be measured using conventional dynamometry. Thus, the calculated tendon forces may have been underestimated.

In the graphical method used for calculating the TA tendon moment arm, the ankle joint was represented as a planar joint model (Maganaris *et al.* 1998*a*, 1999). However, the ankle joint is a three-dimensional mechanism with the tibio-talar and talo-calcaneo-navicular joint axes lying in different anatomical planes (Isman & Inman, 1969).

Consequently foot rotation cannot be performed in a single plane, indicating the possibility of error in the calculation of moment arm.

The TA muscle lies close to the skin and it would be fully activated during percutaneous maximal voltage stimulation. However, in deeply located muscles with motor points away from the skin, percutaneous stimulation would result in only partial muscle activation and current spread into nearby muscles. The anatomical location of the muscle–tendon unit would also affect the analysis of sonographs. Accurate sonograph morphometrics would require high image quality and resolution, and these can be obtained in low-depth scanning only. These effects confine the applicability of the present methodology to superficial muscle–tendon units only.

Major assumptions made in previous studies on *in vivo* human tendon mechanical properties raise serious doubts about the validity of the results obtained. Cook & McDonagh (1996) adapted the *in situ* method of Morgan (1977) in *in vivo* conditions. They separated the muscle–tendon complex into active and passive components and estimated the stiffness of human first dorsal interosseous tendon–aponeurosis, assuming a linear load–elongation relationship, despite experimental evidence of curvilinear increases in tendon force and stress at low strains, in the so-called 'toe' region (Viidik, 1973; Butler *et al.* 1978). More recently Ito *et al.* (1998) estimated the elongation of the human TA tendon during dorsiflexion contraction from the displacement of muscle fasciculi insertion on the aponeurosis, without taking account of changes in the TA tendon orientation upon dorsiflexion loading. This approach would result in valid tendon elongation values only if the aponeurosis and the retinaculum were rigid, which is clearly not the case (Lieber *et al.* 1991; Maganaris *et al.* 1999). Furthermore, and in contrast to our electrical stimulation-based measurements, in the study by Ito *et al.* (1998) stiffness was calculated from tendon forces during isometric voluntary contraction neglecting antagonistic co-activation, which would result in a decreased contraction efficiency due to (a) the negative antagonistic joint moment generated and (b) agonistic reciprocal inhibition (Tyler & Hutton, 1989). Antagonistic co-activation may decrease agonistic joint moment production by up to 35 % (Maganaris *et al.* 1998*b*), indicating that enormous errors may have been incorporated by Ito *et al.* (1998) in the calculation of tendon force and stiffness from net joint moment.

In the present study, tendon stress at maximum isometric contraction intensity was 25 MPa. Since tendons are in series with muscles, the tendon maximum stress equals the muscle specific tension multiplied by the ratio of muscle to tendon cross-sectional areas (Ker *et al.* 1988). By dividing the TA muscle volume, which was calculated from the MRI scans taken according to Fukunaga *et al.* (1992), by previous *in vivo* measures of TA muscle fibre length (Maganaris & Baltzopoulos, 1999), we estimated that the TA muscle cross-sectional area was ~20 cm^2. Assuming that the muscle specific tension is 0·3 MPa (Ker *et al.* 1988) it was calculated that the stress-generating potential of TA tendon would be

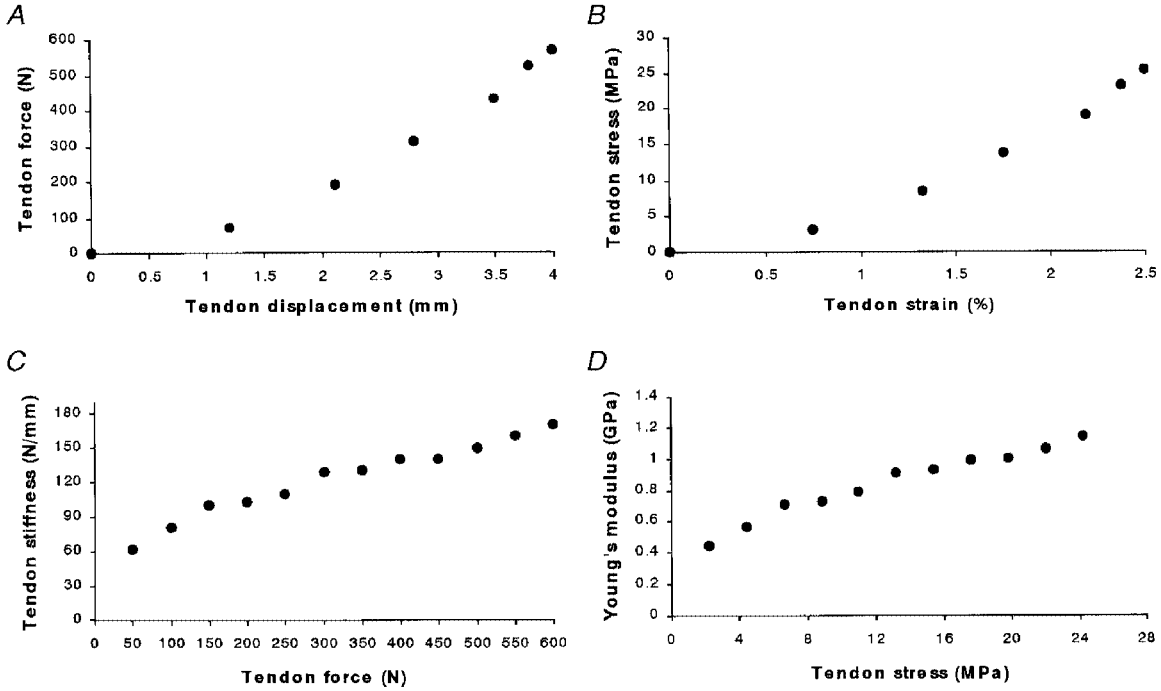

Figure 4. Tendon mechanical properties

A, tendon force–displacement relationship; *B*, stress–strain relationship; *C*, stiffness–force relationship; *D*, Young's modulus–stress relationship. Data from one subject are presented.

~30 MPa, a value close to that measured in the present study. The tendon Young's modulus at maximum tendon stress was 1·2 GPa, which is in line with previous reports (Ker *et al.* 1988; Zajac, 1989). The tendon strain of 2·5% at maximum tendon stress is also in agreement with previous corresponding measurements in isolated tendons (Ker *et al.* 1988; Lieber *et al.* 1991; Loren & Lieber, 1995), but substantially smaller than the value of over 4·3% extracted from the results reported by Ito *et al.* (1998), who included in their measurements the tendon, the aponeurosis and the retinaculum. Considering, however, that (a) aponeuroses are more compliant than tendons (Lieber *et al.* 1991) and (b) stretch of the retinaculum upon loading would result in overestimation of the tendon–aponeurosis elongation, the inconsistency between the findings of Ito *et al.* (1998) and those of the present and other reports would not be unexpected. The maximum strain in the present study was below the higher 'toe' region limit of ~4% (Alexander, 1981; Zajac, 1989), and this indicates that over the range of loads examined the TA tendon operated within the 'toe' region only. This is supported by the fact that our tendon load–displacement and stress–strain curves failed to show 'linear' regions as revealed traditionally in ultimate tensile strength measurements above ~4% strains (Butler *et al.* 1978; Bennett *et al.* 1986). Most everyday life and sporting activities would not require application of dorsiflexion TA forces higher than those elicited during maximal voltage stimulation in our study. This would indicate that the physiological *in vivo* loading of the human TA tendon lies within the 'toe' region. This suggestion is in line with the general observation that most human tendons are not normally at risk of fracture, unless rapid tendon stretch is applied during eccentric muscle contraction (Ker *et al.* 1988). By dividing the average tendon fracture stress of 100 MPa by the maximum tendon stress in the study it was calculated that the TA tendon has a safety factor of 4, which is in line with the safety factors of other human lower limb tendons (Ker *et al.* 1988).

In the present study, the mechanical properties of a human tendon were measured *in vivo*. The measurements taken were reproducible and led to estimations consistent with results from *in vitro* dynamic tensile testing. However, further studies are required to confirm these findings.

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