LOAD–ELONGATION CHARACTERISTICS OF IN VIVO HUMAN TENDON AND APONEUROSIS

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Summary

In the present study, we measured the in vivo load–elongation characteristics of the human tibialis anterior tendon and its central aponeurosis. Measurements were taken in five men using dynamometry, muscle electrical stimulation and ultrasonography. Percutaneous tetanic stimulation of the muscle at successive voltages corresponding to 20, 40, 60, 80 and 100 % of maximum isometric dorsiflexion moment was applied. During electrical stimulation, we recorded the displacements of the tibialis anterior tendon origin and its aponeurosis proximal end using B-mode ultrasonography. Aponeurosis displacement was calculated by subtracting tendon displacement from the displacement of the aponeurosis proximal end. Tendon and aponeurosis displacements increased curvilinearly from 1.3 to 4 mm and from 3.7 to 12 mm, respectively, as a function of dorsiflexion load. Scaling of the displacements recorded to the resting lengths (measured over the skin) yielded strain values that increased curvilinearly with load, from 0.8 to 2.5 % in the tendon and from 2.1 to 7 % in the aponeurosis. Tendon strain was smaller by between 61 and 64 % compared with aponeurosis strain at any given contraction level. These findings are in line with reports from in vitro isolated material testing and have important implications for muscle modelling.

Key words: tendon, aponeurosis, ultrasonography, electrical stimulation, load–elongation, human.

Introduction

Tendinous tissue elasticity has long been studied using one of the following four methodologies: (i) the alpha method (Morgan, 1977), (ii) the spindle null-point method (Rack and Westbury, 1984), (iii) the free vibration method (e.g. Shorten, 1987) and (iv) tensile testing methodologies (e.g. Bennett et al., 1986). Each of these methods has advantages and limitations. The alpha method is adaptable under in vivo conditions (Cook and McDonagh, 1996) but assumes a linear stiffness increase with force and takes account of the whole tendinous component, i.e. extramuscular tendon and aponeurosis. The latter is also the case for the free vibration in vivo method and the spindle null-point in situ method. In tensile testing methodologies, the extra- and intramuscular tendon portions can be tested separately, but preserved or deep-frozen material that may have altered properties (Smith et al., 1996) has traditionally been used (e.g. Bennett et al., 1986).

B-mode ultrasonography offers the possibility for realistic in vivo assessment of human tendinous tissue elasticity. It results in accurate scanning of intramuscular collagenous tissue (Kawakami et al., 1993; Narici et al., 1996) and allows the recording of its movement in intact muscle during contraction (Fukunaga et al., 1996). Selection of appropriate reference points in the extra- and intramuscular tendon portions scanned in combination with artificial selective muscle activation would allow quantification of tendon and aponeurosis displacements upon control loading. In the present study, we combined these techniques to estimate the in vivo load–elongation characteristics of the human tibialis anterior tendon and its central aponeurosis. Our hypothesis was that, for a given load application, tendon elongation would be smaller than that of the aponeurosis.

Materials and methods

Experimental protocol

Measurements were taken in five healthy male volunteers (age 22±4 years; height 172±4 cm; body mass 73±7 kg; lower leg length 40±3 cm, means ± s.d.). The study was approved by the local ethics committee. Isometric ankle dorsiflexion moments were measured in the prone position on an isokinetic dynamometer (Lido Active, Loredan Biomedical, Davis, California, USA) with the knee of the tested leg (the right leg in all subjects) flexed at 90°. The ankle joint axis was visually aligned with the dynamometer axis, and the foot was fixed at the neutral anatomical position (the sole of the foot at 90° to the tibia). At that ankle angle, no passive forces were detected.
by the dynamometer (see also Siegler et al., 1984). To maintain
the ankle joint and dynamometer axes alignment during
dorsiflexion contraction, we bandaged the foot and the heel
around the dynamometer footplate using Velcro straps, and we
placed a mechanical stop below the knee. This system
prevented any observable movement of the lower extremity
during maximum voluntary isometric dorsiflexion. The
dorsiflexion forces elicited in our experiment were much
smaller than those generated upon maximum voluntary effort
(see below). We therefore assumed that any foot movement in
our measurements would have been negligible.
Contraction forces were elicited by means of percutaneous
electrical stimulation of the tibialis anterior muscle using a
custom-built, high-voltage stimulator. A train of bipolar wave
pulses with a duration of 100 μs was applied at a frequency of
100 Hz for 1 s through two 4 cm × 3 cm aluminium foil pads
placed over the motor point area of the muscle. We first
identified the voltage eliciting maximal joint moment ($M_{\text{max}}$).
Maximality was indicated by no further increase in the joint
moment recorded with increasing voltage. We then identified
the voltages eliciting joint moments corresponding to 20, 40,
60 and 80% of $M_{\text{max}}$. Surface electromyographic (EMG)
signals from the nearby soleus and peroneous tertius muscles
showed no evidence of current leakage during stimulation of
the tibialis anterior muscle.
A 7.5 MHz linear-array B-mode ultrasound probe (Esaote
Biomedica, Florence, Italy; width resolution 1 mm, depth
resolution 0.62 mm) was secured with adhesive tape onto the
skin, first over the tibialis anterior myotendinous junction
region, approximately 7 cm above the malleoli, and then over
the proximal region of the tibialis anterior muscle,
approximately 7 cm below the knee. In scans recorded at rest,
we identified the tibialis anterior tendon origin and its central
aponeurosis proximal end. Muscle contractions were then
elicited at successive voltages corresponding to 20, 40, 60, 80
and 100% of $M_{\text{max}}$. The displacements of the tibialis anterior
tendon origin and aponeurosis proximal end in the transition
from rest to all contraction level joint moments were recorded
digitized. The tibialis anterior tendon origin displacement
was considered as the tibialis anterior tendon displacement,
and the aponeurosis proximal end displacement was
considered as the tendon/aponeurosis complex displacement.
Aponeurosis displacement was estimated by subtracting the
tendon displacement from the tendon/aponeurosis complex
displacement (Fig. 1). From each site, we recorded
displacement data from two series of measurements, 2 min
apart, and we used average values for further analysis.
Measurements were taken after the tibialis anterior tendinous
structures had been preconditioned by five series of stimulated
contractions at 20, 40, 60, 80 and 100% of $M_{\text{max}}$. 30 s apart
(see also Bennett et al., 1986). To calculate strain values from
the displacements obtained, we measured the resting lengths of
the tibialis anterior tendon and aponeurosis to the nearest
millimetre over the skin as described by Ito et al. (1998).
Tendon length was estimated as the length from the tibialis
anterior tendon origin to its insertion point on the base of the
first metatarsal bone following the curved path of the tendon.
Aponeurosis length was assumed to be the distance from the
tendon origin to the aponeurosis proximal end (Fig. 1).
To minimize overestimations in the displacements obtained
caused by stretch in the surrounding retinaculum upon
dorsiflexion loading (Maganaris et al., 1999), we carried out
our measurements with the ankle joint bandaged tightly with
inelastic tape. Pilot sagittal-plane magnetic resonance imaging

![Fig. 1. Illustration of the displacements and lengths measured. P and D are the proximal and distal points,
respectively, of the tibialis anterior (TA) muscle/tendon unit. $L_{\text{Tend}}$ and $L_{\text{Apon}}$ are the resting lengths of the tibialis
anterior tendon and its central aponeurosis, respectively. Arrows A and B point to the origin of the tibialis anterior
tendon and the proximal end of the central aponeurosis, respectively. $d_{\text{Tend}}$ and $d_{\text{Total}}$ are the displacements of the
 Tibialis anterior tendon and tendon/aponeurosis complex, respectively, in the transition from rest (top) to
dorsiflexion contraction (bottom). Aponeurosis displacement was estimated by subtracting $d_{\text{Tend}}$ from
$d_{\text{Total}}$.](image-url)
In vivo human tendinous tissue elongation

(MRI)-based morphometrics in the ankle joint indicated that this fixation system was effective in maintaining the resting curved path of the tibialis anterior tendon during maximum isometric voluntary dorsiflexion.

All measurements were taken 4 days after a familiarization trial. All analyses were performed three times by the same investigator, and average values were used for further analysis.

Statistical analyses

Values are presented as means ± S.D. Friedman’s rank test was used to test differences in the tendon and aponeurosis elongations as a function of contraction level joint moment. The Mann–Whitney test was used to test for differences between the tendon and aponeurosis elongations at a given contraction level joint moment. Statistical difference was set at a level of \( P < 0.05 \).

Results

Maximal-voltage stimulation of the tibialis anterior muscle yielded joint moment values corresponding to approximately 45% of the moment generated upon maximum voluntary isometric dorsiflexion. In the transition from rest to \( M_{\text{max}} \), both the tibialis anterior tendon origin and the central aponeurosis proximal end shifted proximally (Fig. 2). As the dorsiflexion joint moment increased from 20 to 100% of \( M_{\text{max}} \), the displacement of the tibialis anterior tendon origin at rest increased curvilinearly from 1.3±0.4 to 4±1.3 mm \((P<0.01)\) (Fig. 3A). The calculated aponeurosis displacement increased also curvilinearly from 3.7±1.3 at 20% of \( M_{\text{max}} \) to 12±4 mm \((P<0.01)\) at 100% of \( M_{\text{max}} \). Tendon displacement was smaller by between 2.4 and 8 mm \((P<0.01)\) compared with that of the aponeurosis at any given contraction level joint moment. The resting lengths of the tibialis anterior tendon and central aponeurosis were 160±16 and 170±20 mm, respectively. The load–strain relationships of the tibialis anterior tendon and its aponeurosis were also curvilinear (Fig. 3B). Strain increased as a function of load from 0.8±0.1 to 2.5±0.4% \((P<0.01)\) in the tendon and from 2.1±0.4 to 7±1.3% \((P<0.01)\) in the aponeurosis. Tendon strain was smaller by between 61 and 64% \((P<0.01)\) compared with that of the aponeurosis at any given contraction level joint moment (Fig. 3).

Reproducibility of measurements

The displacements of the tibialis anterior tendon origin and aponeurosis proximal end in the transition from rest to 40% of \( M_{\text{max}} \) were recorded and digitized in one subject on 10 occasions after the tendon had been preconditioned (see Fig. 2. Typical sonographs at rest (top), at 40% of maximal joint moment \( (M_{\text{max}}) \) (middle) and at 100% of \( M_{\text{max}} \) (bottom). The white arrows point to the tibialis anterior tendon origin (A) and the proximal end of the tibialis anterior central aponeurosis (B). The black arrows point to the shadow generated by an echo-absorptive marker attached with adhesive on the skin to confirm the constancy of the scanning probe during contraction.
The coefficients of variation for the repeat scans were 3.3 and 5.2 %, respectively. Four different observers recorded and analyzed the displacements of the tibialis anterior tendon origin and aponeurosis proximal end in the transition from rest to 20 % of $M_{max}$ in a given subject, and the coefficients of variation were 6.1 and 8.4 %, respectively. Measurements of the resting lengths of the tibialis anterior tendon and aponeurosis were made 15 times and resulted in coefficients of variation of 3.7 and 1.9 %, respectively.

**Discussion**

In the present study, we have demonstrated (i) that, for given loads, the intact human tibialis anterior tendon lengthened less than its central aponeurosis and (ii) that the tibialis anterior tendon and aponeurosis load–elongation relationships in vivo were curvilinear up to $M_{max}$. Both these findings are in line with previous reports from in vitro tensile testing of isolated material (Butler et al., 1978; Bennett et al., 1986; Huijing and Ettema, 1988/1989; Ettema and Huijing, 1989; Trestik and Lieber, 1993). Inter-study differences in the methodologies followed may account for the disagreement in the results obtained.

In our study, the aponeurosis was examined under physiological conditions, attached to the muscle (see also Zuurbier and Huijing, 1992; van Donkelaar et al., 1999), and it lengthened upon muscle contraction. This finding has important implications for muscle modelling applications. In situ and in vivo experiments have demonstrated differences between fibre and muscle length decreases during contraction (Woittiez et al., 1983; Griffiths, 1991). Pennate muscle models incorporating inextensible aponeuroses have often failed to predict accurately the actual behaviour of muscle fibres during contraction, thus introducing errors in the analysis of forces (Huijing and Woittiez, 1985; Maganaris et al., 1998). The discrepancy in strain between the tendon and aponeurosis may partly account for the intrinsic material compliance are probably the cause of the strain difference between the tendon and aponeurosis (Rack et al., 1984; Scott and Loeb, 1995). However, some studies have reported similar strain values for the tendon and aponeurosis (Trestik and Lieber, 1993). However, some studies have reported similar strain values for the tendon and aponeurosis (Trestik and Lieber, 1993). Inter-study differences in the methodologies followed may account for the disagreement in the results obtained.

In our experiment, the tibialis anterior tendon exhibited a strain of 2.5 % at $M_{max}$, while its central aponeurosis experienced a strain of 7 %, almost three times as much as that of the tendon. Several authors have reported values similar to our strain values for animal and human tendons and aponeuroses at loads equivalent to maximal muscle force (Ker et al., 1988; Huijing and Ettema, 1988/1989; Ettema and Huijing, 1989; Lieber et al., 1991; Loren and Lieber, 1995). The discrepancy in strain between the tendon and aponeurosis is consistent across most of the studies and indicates a decreased stiffness in the aponeurosis compared with the tendon. Experimental evidence indicates that differences in specimen cross-sectional area rather than in intrinsic material compliance are probably the cause of the strain difference between the tendon and aponeurosis (Zuurbier and Westbury, 1984; Scott and Loeb, 1995). However, some studies have reported similar strain values for the tendon and aponeurosis (Trestik and Lieber, 1993). Inter-study differences in the methodologies followed may account for the disagreement in the results obtained.

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100% of $M_{\text{max}}$. From the slope of the tendon tension–displacement curve, we calculated that tendon stiffness would increase from 60 to 160 N mm$^{-1}$. Aponeurosis tension and stiffness, however, cannot be estimated from our data. In the model shown in Fig. 1, the contractile force by which the aponeurosis proximal end is pulled is smaller than the force exerted on the tendon origin. It must be realized, however, that such muscle models are oversimplified. They assume (i) a straight-line orientation between the tendon and the aponeurosis and (ii) that all muscle fibres are in a perfectly parallel arrangement with no transverse inter-fibre connections that would result in lateral force transmission. Experimental evidence has shown that both these assumptions are invalid (van Leeuwen and Spoor, 1992; Trotter, 1993), indicating that erroneous force estimations could be derived using simple geometric models.

We observed large variations in the elongations of the tendon and aponeurosis in our subjects (see Fig. 3). On average, strain values for a given load were larger by approximately 35% in sedentary subjects ($N=3$) than in subjects undergoing physical training ($N=2$). Although our small sample size does not allow any conclusions to be drawn, the observation of stiffer tendinous structures in response to exercise training is consistent with previous reports from animal studies (for a review, see Tipton et al., 1986). This indicates that the present methodology may be sensitive enough to detect exercise- and disuse-induced changes in intact human tendinous structure elasticity, possibilities that are currently being investigated systematically.

References


