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Creep and the in vivo assessment of human patellar tendon mechanical properties

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Abstract

Background. Owing to the viscoelastic nature of tendons it may be that the total excursion and hence strain experienced by the tendon under load may be affected by the duration of contraction. Here we examine the effect of contraction duration on the measured in vivo mechanical properties of the patellar tendon.

Methods. Nine healthy young men aged 21 (SEM 0.5 years) performed three short (3–4 s) and three long (10–12 s) maximal ramped isometric contractions on an isokinetic dynamometer, with real-time recordings of patellar excursions using B-mode ultrasonography synchronised with forces to determine tendon mechanical properties.

Findings. Maximal patellar excursion was ~42% ($P < 0.001$, effect size (r) = 0.9) lower for the short 3.6 (SEM 0.4 mm) vs. the long 6.2 (SEM 0.4 mm) contractions. Similarly, across the range of forces tested, strain was ~42% ($P < 0.001$, $r = 0.9$) lower for the short vs. the long contractions 4.5 (SEM 0.5) vs. 8.0 (SEM 0.9%), respectively. Tendon stiffness however, was ~77% greater (4648 SEM 434 vs. 2633 SEM 257 N mm^{-1}, $P < 0.001$, $r = 0.9$) for short vs. long contractions.

Interpretation. Contraction duration significantly affects tendon strain and associated measures of stiffness at all levels of force. The implications of this finding are twofold in that the results: (a) indicate that in order to compare tendon mechanical properties within or across studies, duration of contraction must be standardised and (b) have possible implications on training protocols and associated injury risks.

Keywords: Creep; Patella tendon; Viscoelasticity

1. Introduction

During muscle contraction, forces are transferred from the muscle to the bone via tendons. In particular, the quadriceps muscle group transfers forces to the lower limb via the patellar tendon, which is attached at the patella and inserts at the tibial tuberosity. At the cellular level, the tendon is mainly composed of type I collagen fibres which possess certain mechanical properties, namely that they are generally considered as responding to stresses in a viscoelastic manner.

Thus, the tendon deforms in a non-linear manner in response to the applied loading characteristics (Onambele-Pearson and Pearson, 2006; Pearson and Onambele, 2005, 2006). More specifically, tendinous tissue deformation develops in a time dependant characteristic manner in response to a constant load (see Fig. 1a and b). Furthermore, tendinous tissue deformation is also affected both by the material stiffness of the tendon itself, but also by the absolute degree of loading i.e. the amount of force (Maganaris and Paul, 1999). Previous authors (e.g. Maganaris et al., 2002) have reported that during consecutive contractions, medial gastrocnemius tendon strain, at the same levels of force, increased. This would tend to suggest that in vivo creep in human tendons may be an important factor to consider where the nature of the effects of a loading protocol is being studied. It should be noted here that the ''Creep'' terminology is considered to be a general characteristic of viscoelastic materials to undergo increased deformation under a constant stress, until an asymptotic level of strain is reached (see Fig. 1b).

Another important characteristic of tendons is stiffness. Stiffness is a property of the tendon which is principally associated with the unit's ability to transfer forces, a stiffer tendon being able to transfer muscle forces to the bone more rapidly than a less stiff (or more compliant) tendon. Also, a tendon can act like a shock absorber whereby possessing low stiffness can lead to the damping of high and potentially injury-inducing forces for example when landing in running or jumping. Hence in this example, low tendon stiffness might possibly help in reducing any potentially damaging high impact forces.

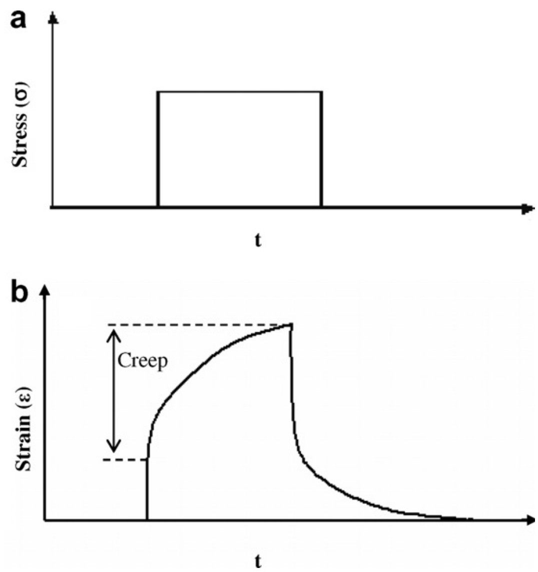

Fig. 1. Generic expectations from the behaviour of viscoelastic materials:
(a) Step loading protocol with characteristic of constant stress and
(b) relationship between duration of contraction and non linear creep. Data are theoretical models. For further details relating to the description of creep in soft collagenous tissue, please refer to a model by Vena et al. (2006).

Numerous earlier studies have utilised ultrasound to determine in vivo tendon mechanical properties by recording the deformation of the tendon via real-time monitoring of patellar excursions under ramped isometric loading conditions (Onambele-Pearson and Pearson, 2006; Pearson and Onambele, 2005, 2006; Reeves et al., 2003). More recently Hansen et al. (2006) have also measured patellar tendon mechanical properties under similar loading conditions. However, these authors utilised a ramped period of force which was substantially longer (~10 s) than that used in previous studies, where a typical contraction time of 3–4 s has been used. This extended duration of force application could potentially have an effect on the measured in vivo tendon deformation owing to the characteristics of creep. No agreed standard exists for the technique of in vivo assessment of tendon mechanical properties, particularly where the duration of force application is concerned. If it can be shown that duration of force application has an effect on the measured tendon deformation it would be pertinent for all future studies which examine tendon mechanical properties to adopt a standard protocol to allow comparisons between studies.

With the above in mind, the aim of the present study was to examine the in vivo effect of duration of contraction on the measured strain and stiffness of the patellar tendon. Our hypothesis being that as duration of force application increases, the observed strain of the tendon at a given force will increase; hence stiffness will be seen to decrease.

2. Methods

2.1. Subjects and experimental design

Nine males aged 21 (SEM 0.5 years), with a mass of 88.2 (SEM 3.7 kg) and a height of 183.4 (SEM 2.1 cm), volunteered to take part in the study. All subjects gave their writ- ten informed consent to participate in this study. The study had the approval of the local University Ethics Committee and was in agreement with the World Medical Association's declaration of Helsinki describing ethical principles for medical research involving human subjects. All participants had a familiarisation session in the laboratory before any testing was carried out. All tests were carried out after a standardised warm up (3 maximal isometric knee extension and flexion contractions at 90° knee angle (with full knee extension = 0°)) on an isokinetic dynamometer (Kin Com dynamometer, type 125 AP, Chattanooga, USA).

2.2. Measurement of patellar tendon forces

As shown in Fig. 2, all knee extension exertions were carried out on the same dynamometer used during the familiarisation and warm up protocols. Subjects were instructed to develop three maximal voluntary isometric efforts over either 3–4 s (P_{short}) or 10–12 s (P_{long}) in a randomised order with 180 s rest between subsequent contractions.

Tendon force was calculated as $F_{tend} = (P + P_{antag})/\ T_{arm}$ where F_{tend} = force in the patellar tendon, P = observed knee extensor torque output, P_{antag} = antagonis- tic (hamstring) co-contraction torque, and T_{arm} = patellar tendon moment arm 44.7 mm (Lu and O'Connor, 1996).

Fig. 2. Experimental set-up. Isometric knee extension torque was deter- mined by fixing the knee at 90° flexion (full extension = 0°) and hip at 85° (supine = 0°). The knee joint centre was aligned with the centre of rotation of the dynamometer lever arm and straps were fixed across the chest, hip and thigh of the test limb to prevent any extraneous movement. A padded cuff was strapped on the lower leg at ~3 cm above the medial malleolus. Inset - showing sagittal plane ultrasound image of patellar and tendon.

2.3. Estimation of Hamstring co-contraction

Hamstring torque was estimated by using electromyography (EMG). The EMG of the long head of the BF was measured in order to ascertain the level of antagonistic muscle co-contraction (Onambele-Pearson and Pearson, 2006; Pearson and Onambele, 2005, 2006) during the isometric knee extension performances. Assumptions were that BF is representative of its constituent muscle group (Carolan and Cafarelli, 1992), and BF EMG relationship with knee flexors torque is linear (Lippold, 1952; Reeves et al., 2003).

A pair of self-adhesive Ag–AgCl electrodes ~15 mm in diameter (Medicotest, Rugmarken, Denmark, type N-10A), was placed in a bipolar configuration with a constant between-electrodes distance of ~20 mm, at the distal one- third of the length, in the mid-line of the belly of the biceps femoris muscle (BF). Prior to electrode attachment the skin was prepared by shaving, abrading, and cleaning with an alcohol-based solution in order to minimise the resistance. The reference electrode (Type Q-10A) was placed on the lateral tibial condyle of the test limb. The raw EMG signal was sampled at 2000 Hz, pre-amplified (x2000, Neurolog remote AC preamplifier, type NL 822, Digitimer, UK), amplified (x2) (Neurolog isolation amplifier NL 820, Digitimer, UK) and band pass filtered between 500 Hz and 10 Hz (Neurolog, type NL 134 and NL 144, Digitimer, UK). All EMG and torque signals were displayed in real time in Testpoint software (CEC, MA, USA) via a PC. A series of 4 maximal flexion contractions were carried out to obtain the EMG value 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 (SENIAM, 1999). Electromyographic activity of the BF during knee extension was divided by the maximal BF flexor EMG, and the maximal flexor torque was then multiplied by this value to determine co-contraction torque.

2.4. Tendon mechanical properties

Tendon force–elongation relationships were determined, as described previously (Pearson and Onambele, 2006). Briefly, simultaneous recording of patellar excursion, generated torque and EMG were carried out during the ramped isometric knee extensions. Ultrasound images were generated using B-mode ultrasound (AU5, Esaote, Genoa, Italy) with a 7.5 MHz linear array probe (38 mm wide) aligned in the sagittal plane over the patellar tendon. Tendon images were digitised directly to a PC at 25 frames s^{-1} (Adobe Premier pro Ver.2) and subsequently analysed using image J (National Institute of Health, Bethesda, MD, USA). All signals were synchronised to allow temporal alignment.

Tendon strain (%) was calculated as the ratio of tendon deformation to the resting tendon length (T_L) i.e. the distance between the apex of the patella and the insertion point at the tibial tuberosity. Plotted force–elongation data were fitted with a second order polynomial function forced through zero to allow the calculation of tendon stiffness.

Stiffness values (K in N mm^{-1}) or the gradients of the tangents to the polynomial functions, were then computed at force levels corresponding to the group's average values at 10%, 30%, 40%, 50%, 60%, 70%, 90% and 100% of maximal voluntary force (see Table 1).

2.5. Statistical analyses

Tests for differences between the means for (P_{short} and P_{long}) of the calculated mechanical parameters were determined using paired Student *t*-tests. Significance was set to $P < 0.05$. Intraclass correlation coefficients (one way random effects model) were calculated to estimate reliability of the measures. Effect size (r) was calculated for all paired *t*-tests.

Table 1
Effect of creep on the assessment of in vivo mechanical properties of human patellar tendon

Force (%)	Force (N)	Slow ramped contractions			Fast ramped contractions		
		Deformation (mm)	Strain (%)	Stiffness (N mm^{-1})	Deformation (mm)	Strain (%)	Stiffness (N mm^{-1})
~10	1136	2.0 (0.3)	3.6 (0.5)	1344 (120)	1.0 (0.2)	1.8 (0.2)	2473 ± 373
~30	3313	3.3 (0.3)	6.0 (0.7)	2057 (152)	1.8 (0.2)	3.3 (0.3)	3668 ± 511
~40	4388	3.8 (0.3)	7.0 (0.7)	2328 (169)	2.2 (0.2)	3.8 (0.4)	4128 ± 573
~50	5496	4.3 (0.3)	7.8 (0.8)	2578 (186)	2.4 (0.3)	4.4 (0.4)	4549 ± 632
~60	6600	4.7 (0.4)	8.6 (0.8)	2804 (201)	2.7 (0.3)	4.9 (0.5)	4934 ± 686
~70	7671	5.1 (0.4)	9.3 (0.9)	3007 (215)	2.9 (0.3)	5.3 (0.5)	5279 ± 735
~90	9886	5.9 (0.4)	10.6 (1.0)	3389 (242)	3.4 (0.3)	6.1 (0.6)	5930 ± 829
~100	10960	6.2 (0.4)	11.2 (1.0)	3559 (254)	3.6 (0.4)	6.5 (0.7)	6221 ± 871
Mean across all force levels		4.4 (0.5)	8.0 (0.9)	2633 (257)	2.5 (0.3)	4.5 (0.5)	4648 ± 434

Force (%) represents the closest round number to the relative force level at which individual data was calculated. Force (N) represents the absolute group mean data corresponding to these relative force levels. Thus all subjects' elongation, strain and stiffness were calculated at the same absolute values to allow direct comparison between conditions. All values are mean (SEM).

3. Results

The within session ICCs for the long duration contractions were 0.996 and 0.925 for tendon force and elongation respectively ($P < 0.001$). The ICCs for the short duration contractions were 0.978 and 0.982 for tendon force and elongation respectively ($P < 0.001$). It is notable that although the study subjects were very repeatable within themselves, they consistently produced greater maximal voluntary forces in the fast (~3 s) compared with the slow (~10 s) contractions 13268 (SEM 1008 N) vs. 10 960 (SEM 965 N), ($P < 0.001$), an effect which was not due to fatigue since the order of contraction rates in the test protocol was randomised. This observation of lower maximal forces in one protocol however, did not affect the interpretation of our data since the comparisons of tendon properties discussed below (also see Table 1) were taken (unless otherwise specified) at the same force levels for both the fast and the slow contraction.

The force–elongation graph (Fig. 3) illustrates the differences in patellar excursion for the P_{short} and P_{long} protocols. It is clear that at similar levels of force experienced by the tendon during the ramped contractions, a greater elongation is observed for the P_{long} protocol. On comparison of maximal tendon elongation values, maximal tendon elongation for P_{short} was 3.6 (SEM 0.4 mm) compared with 6.2 (SEM 0.4 mm) for the P_{long} efforts. This represents a significant 42% ($P < 0.001$ effect size (r) = 0.9) lower maximal tendon deformation during the short ramped contractions in contrast to that during the long ramped contractions. It is also notable that level of force did not have a significant effect on the protocol-induced error in strain measurement with the average of the pair wise differences in slow versus fast strains remaining constant between ~38% and 43% (see Table 1).

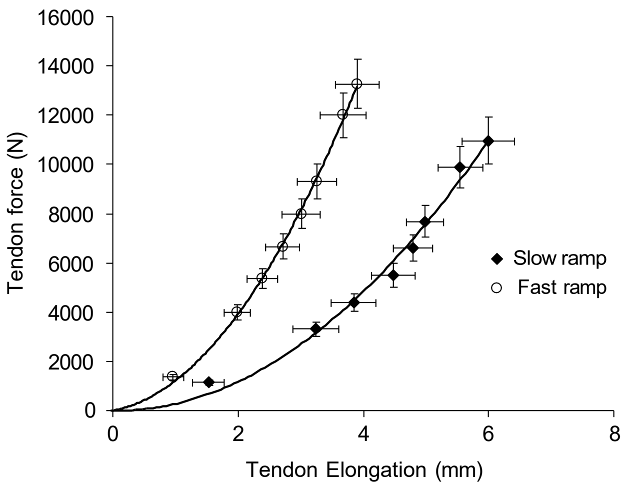

Fig. 3. Tendon force-deformation curve. Closed symbols represent slow (~10 s), and open symbols represent fast (~3 s) measurements. At all force levels, deformation in the slow, was consistently and significantly increased (~38–43%, $P < 0.001$), compared to values in the fast protocol. Data are mean (SEM).

Stiffness values calculated from the force elongation data (see methods for detail) were consistently different across the absolute force levels identified, with the short ramped contractions giving ~77% greater values of stiffness across the range (Table 1). Group stiffness values at forces ranging from ~1200–11000 N were shown to be significantly different ($P < 0.001$, $r = 0.9$) with a mean of 2633 (SEM 257) and 4648 (SEM 434 N mm^{-1}) for P_{long} and P_{short}, respectively.

4. Discussion

This study examined the effects of contraction duration on the in vivo measurement of tendon mechanical properties. Specifically, whether/how contraction duration would affect patellar excursion and hence calculated strain and stiffness. The hypothesis being that, owing to the known in vitro visco-elastic properties of tendon materials, an increased duration of force application in vivo would lead to a significant increase in the tendon elongation and hence reduction in calculated tendon stiffness. Our study findings have confirmed this hypothesis.

4.1. Tendon viscoelastic properties and creep

By examination of the characteristic time-extension curve derived during tendon loading over time it can be seen that if a load is applied and the tendon is given a finite time to develop strain, the extent to which the tendon elongates will be dependent upon the time at which the measurement is made (Fig. 1b). This effect is termed tendon creep; the most significant effect is suggested to occur during the first few contractions (Schatzmann et al., 1998).

Based on the above assumption, it is common practise to carry out ''preconditioning'' efforts prior to any measurement of the tendon properties. Previous studies have utilised four preconditioning trials prior to measurement of tendon properties (Kubo et al., 2002; Reeves et al., 2003; Onambele-Pearson and Pearson, 2006). However, it may be that in spite of these preconditioning efforts, the tendon may still possess the ability to express creep. In fact it has been suggested that creep is observed in tendon for approximately 160 contractions (Schatzmann et al., 1998). Others have also noted creep behaviour in tendon after preconditioning. De Zee et al. (2000) utilised a pre-conditioning protocol of 1.5 Hz (using a haversine pattern) for one minute with loads of 100–350 N. The subsequent loading with a constant force (to a level of 3% strain) showed a non linear increase in strain, reaching a plateau after 1 min. It was clear from the results that significant creep was exhibited within the 10 s time frame as used in the current, and in a previous study (Hansen et al., 2006). The observed levels of maximal strain in this study for the short contractions 6.5 (SEM 0.7%) are slightly lower than others that have measured the patella tendon (9.9%) (Reeves et al., 2003), however the subjects used by Reeves were elderly who have been shown to have less stiff tendons compared with their younger counterparts (Onambele et al., 2006). The values for the longer contractions here were very similar to those of Reeves et al. (11% vs. 9.9%), but here different contraction times were used making comparisons difficult.

4.2. Duration of loading and calculated mechanical properties

Owing to the different loading durations used in the current study, strain was affected, ultimately changing the calculated measures of stiffness. Values of calculated stiffness at absolute levels of force were seen to be significantly different between the two loading protocols with all values significantly lower for the longer contractions. Group mean values of stiffness observed at the shorter contraction efforts were seen to be approximately twice that of others who have also used short contractions (Reeves et al., 2003), again these were older subjects (mean age ~74 yr). These values are probably reflective of the differences in the collagenous matrix between the young and older subjects.

From our current results, it is clear that tendinous tissue elongation is affected differentially by duration of loading and hence it would seem reasonable for future studies to try and determine the stiffness or strain of the tendon component under conditions which are similar to those in which the tendon will be loaded 'functionally' e.g. cyclically as in gait, or for a predetermined period such as during a sporting action like weight lifting.

4.3. Creep vs. pre-conditioning

There is a distinction between (a) pre-conditioning contractions and (b) viscoelastic creep (whereby a load is applied over an extended period of time: the crux of the current work). Indeed the majority of previous work has only described the effect of (a) i.e. the result that applying a number of contractions at the same strain rate has on the eventual elongation of the tendon. In most cases, including the current study, after approximately the third such contraction, degree of strain stabilises: we call this effect tendon pre-conditioning. The novelty of our current work lays in the fact that what we are demonstrating is that even after taking into account the effects of pre-conditioning, it is the duration of contraction which affects the eventual total tendon elongation. Hence, what we are comparing here is not the effect of loading/unloading a tendon several times. The focus of the current study is the investigation of the difference between the elongation of a tendon, *continuously* loaded over a short period of time, compared with one *continuously* loaded over a relatively extended period of time. Here an approximately threefold difference in strain rate between the two conditions (short vs. fast contractions), resulted in a 42% increment in strain (3). Owing to the uniqueness of our current work, it has generally been inappropriate to make direct comparisons between the current work and previous studies. Indeed Maganaris (2003), for instance studied the effect of repeated contractions on gastrocnemius tendon elongation. This author showed an initial maximal tendon elongation of 15 mm and after a series of loading/unloading, they observed a maximal tendon elongation of 20 mm – this alone would represent an approximate increase of 33% in strain. Blevins et al. (1994) on the other hand used continuous load applications as in our current study, with the crucial difference however that these authors utilised strain rates which were much in excess of those used in our study (in the order of 10–100 · faster). Hence we argue here that in order to observe the effect of the in vivo viscous properties of the human patellar tendon (creep), a much lower rate of loading is required: this is the methodology we have employed in our current study.

4.4. Creep: artefact or physiological event?

It has been demonstrated that some joint movement occurs during isometric contractions thereby impacting on the physiological measures of interest. This was first demonstrated by Magnusson et al. (2001) for the triceps surae, and Bojsen-Moller et al. (2003) found similar mechanisms for the knee extensor tendon complex. In fact, this latter paper reports a knee rotation of up to 10° due to lack of firmness in the dynamometer and associated cushioning. This problem can be addressed by measuring knee joint movement by goniometer (Bojsen-Moller et al., 2003), by kinematic analysis (Stafilidis et al., 2005), or perhaps more directly, by visualizing the entire tendon (Hansen et al., 2006).

Therefore, provided any joint rotation is consistent under both conditions in the present study (i.e. fast vs. slow-ramped contractions), it could be argued that any difference in tendon strain was indeed physiological as opposed to artefactual. Using a goniometer (Biometerics Ltd, Gwent UK), we observed 4.42 (SEM 0.33°) of knee joint rotation during the 3s-protocol, which was not significantly different from the 4.24 (SEM 0.21°) of movement observed during the 10s-protocol (n = 10 trials, P = 0.59). Hence the presently observed differences in mechanical properties are not likely to be due to a difference in the elastic properties of the dynamometer cushioning between fast and slow loading. We acknowledge that we have not accounted for aberrant motions of the tibia during the contraction and as such we have not here shown total creep. Nevertheless, the principal aim, which was to demonstrate in vivo creep, has been achieved.

5. Conclusions

In conclusion, we have shown here that by manipulating the loading duration, different values for the tendon mechanical properties are seen. It is therefore important that for comparison purposes a similar loading protocol is used. The technique for assessing the in vivo characteristics of tendon has potential for uses in prediction of injury risk, rehabilitation and muscle–tendon modelling, to be valid the method needs to be standardised across studies. Additionally, there is scope here for future studies on the effect that differential contraction–duration training may have on the viscoelastic properties of human tendons.

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