Patellar tendon in vivo regional strain with varying knee angle

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ABSTRACT

Purpose: Descriptive data on the aspects of site specific in vivo tendon strain with varying knee joint angle are non-existent. The present study determines and compares surface and deep layer strain of the patellar tendon during isometric contractions across a range of knee joint angles.

Methods: Male participants (age 22.0 ± 3.4) performed ramped isometric knee extensions at knee joint angles of 90^0, 70^0, 50^0 and 30^0 of flexion. Strain patterns of the anterior and posterior regions of the patellar tendon were determined using real-time B-mode ultrasonography at each knee joint angle. Regional strain measures were compared using an automated pixel tracking method.

Results: Strain was seen to be greatest for both the anterior and posterior regions with the knee at 90^0 (7.76 ± 0.89% and 5.06 ± 0.76%). Anterior strain was seen to be significantly greater ($p<0.05$) than posterior strain for all knee angles apart from 30^0, 90^0 = (7.76 vs. 5.06%), 70^0 = (4.77 vs. 3.75%), and 50^0 = (3.74 vs. 2.90%). The relative strain (ratio of anterior to posterior), was greatest with the knee joint angle at 90^0, and decreased as the knee joint angle reduced.

Conclusions: The results from this study indicate that not only are there greater absolute tendon strains with the knee in greater flexion, but that the knee joint angle affects the regional strain differentially, resulting in greater shear between the tendon layers with force application when the knee is in greater degrees of flexion. These results have important implications for rehabilitation and training.

Key Words: Tendon; localised Strain; Patella; Knee extension; Isometric

INTRODUCTION

The tendon structure in humans is described as viscoelastic in nature and is capable of transferring high loads to the bony structures to enable stabilisation, movement and locomotion. Although this structure is strong and generally able to withstand high transient loads, it is at risk of both chronic and acute injury. With this high loading, there is potential for acute injury to the tendon structure due to unaccustomed high loading rates or unstable/unequal loading (i.e. landing from a jump) (Richards et al., 1996). Injuries to the structures around the knee are common in sport, with those sports involving jumping particularly prone to tendon overload injuries (Van der worp et al., 2011; Zwerver et al., 2011; Visnes et al., 2014).

Repeated chronic overload or under load of the tendon could lead to tendinopathic-type changes, resulting in inflammatory or degenerative tendon conditions (Leadbetter 1992; Riley 2008; Rufai et al., 1995; Vogel et al., 1993). In particular, athletes show characteristic issues related to the tendon whereby the structure can either become inflamed or degenerative. These conditions which are thought to be more chronic are termed 'tendinopathies' (Astrom and Rausing, 1995; Kader et al., 2002; Maffulli and Kader, 2002; Regan et al., 1992). A number of factors may be responsible for the development of such conditions, including inappropriate training loads or volumes (Korkia et al., 1994), improper application of technique or equipment (Ilfeld, 1992; James, 1995; Kibler et al., 1992), and imbalance of the contractile and supporting soft tissue system (Almekinders, 1998; Herring and Nilsen, 1987).

With respect to insertional tendinopathies, a biomechanical mechanism has been put forward termed 'stress shielding' (Orchard et al., 2004), whereby the posterior patellar tendon is 'shielded' from the loading experienced by the anterior tendon. Here it is suggested that the 'lack' of loading at the posterior region of the tendon may in fact result in cartilaginous or atrophic type changes over time (Rufai et al., 1995; Vogel et al., 1993). It could also be suggested that as the knee moves through its range of motion, the tendon is differentially exposed to load throughout its cross-section, thus resulting in shear type stressors, causing micro damage which could accumulate to a point where tangible injury to the structure is evident (Riley, 2008). Previous work on both the Achilles and patellar tendons in cadaveric specimens has shown that differential strain exists in these structures when loaded (Almekinders et al., 2002; Basso et al., 2002; Lersch et al., 2012), although these studies do not have consensus on the patterns of differential loading across the tendon. Cadaveric specimens however, do not always represent valid measures for structures in living individuals for a number of reasons including changes in the specimens after death, alterations in the structures due to preparation and testing limitations (slipping of tissues in holding clamps). More recently, using automated non-invasive tracking via ultrasound imaging, it has been reported in humans that there are in vivo differences in regional strain for the patellar tendon during loading (Arndt et al., 2012; Pearson et al., 2014), indicating that shear is present in the tendon structure when loaded in vivo. However, to date there is no data on the potential for differential strain in vivo of the patellar tendon over a range of knee angles, something which occurs normally during movement. As the patellar tendon shows a change in the moment arm across the range of knee angles (Krevolin et al., 2004; Tsaopoulos et al., 2006), it may be suggested that the localised strains seen previously at 90^0 of knee flexion (Pearson et al., 2014) may alter with knee joint angle (See figure 5 for illustration of moment turning effect).

Therefore, the purpose of the current study was to quantify and compare the regional strain patterns (anterior/posterior) of the patellar tendon using a previously described speckle tracking method over a range of knee angles during maximal isometric contractions. It was hypothesised that there would be differences in strain with changing knee angle, and that the ratio of anterior to posterior strain would change with knee angle. The information from this study will enable a more realistic and complete understanding of tendon strain behaviour during loading over a range of joint angles, thus informing the development of rehabilitation, training and injury prevention strategies.

METHODS

Subjects and experimental design:

Ten healthy male participants (aged 22.0 ± 3.4, height 1.75 ± 0.03m, body mass 80 ± 4.3 kg) took part in the cross-sectional study. The investigation was approved by the local ethics committee and all participants gave their written informed consent to take part. The study conformed to the principles of the World Medical Association's Declaration of Helsinki. All participants had a familiarisation session in the laboratory prior to any testing.

Methods

A 7.5 MHz 100mm linear array B mode ultrasound probe (Mylab 70, Esaote Biomedica, Italy) with a depth resolution of 67mm was used to image the patellar tendon in the sagittal plane (ensuring both bone ends were in view - patella and tibia), with the knee joint fixed at a range of flexion angles (90^0, 70^0, 50^0 and 30^0). Ultrasound images were captured at 25Hz for later processing. All scaling in pixels per millimetre was determined from image J software (NIH), by using the known depth of field in the ultrasound images (1mm = 11 pixels in the x and y directions), and used as a calibration factor in the automated tracking programme to ensure equivalent pixel to mm ratios.

Subjects were then required to perform ramped isometric knee extension contractions and simultaneous torque outputs were recorded using a dynamometer (Type 125 AP; Kin Com, Chattanooga, TN) with the participant in a seated position. A series of three ramped contractions (180 seconds rest between each) were carried out at each of the four knee angles (90^0, 70^0, 50^0, 30^0, full extension = 0^0), with the hip fixed at 85^0, (supine = 0^0). Torque was transmitted via a cuff and lever attachment placed at the lower leg ~ 3 cm above the medial malleolus. Leg mass (gravity correction) at all angles apart from 90^0 was calculated and added or subtracted from the output extension or flexion forces respectively (horizontal force (leg mass) x cos ø knee angle), and converted to torque via multiplication of the level arm. Prior to any efforts. three maximal isometric conditioning contractions were carried out. Instructions were given to develop the ramped isometric tension from rest to maximal over a period of 3-4 seconds. The mean value of strain from each series of three contractions at each knee angle were used for later analysis. To enable the torque to be synchronised with the ultrasound output, a square wave signal generator was used.

The electrical activity of the long head of the biceps femoris (BF) muscle was also measured via electromyography (EMG) to ascertain the level of antagonistic muscle co-contraction during the isometric knee extension efforts (Pearson and Onambele, 2006). This was in turn used to adjust the net knee extension torque to calculate total torque (see below). The long head of the BF has previously been shown to be representative of the hamstrings group (Carolan and Cafarelli, 1985) and the relationship between BF torque and EMG reported to be generally linear (Lippold, 1952). In brief, the maximal flexion EMG was determined from a series of three knee flexion efforts. Then the root mean square activity corresponding to the peak torque period was analysed over 50 ms epochs and averaged for 1 sec during the plateau of the isometric maximal effort, this was carried out at each tested knee angle. This has previously been suggested to be acceptable in terms of signal-to-noise ratio (Hermens et al., 1999). Any subsequent EMG of the BF during maximal knee extension efforts was divided by the maximal flexor torque EMG, and the maximal flexor torque was then multiplied by this value to determine co-contraction torque. Patellar tendon forces were then determined by dividing the total torque by the patellar tendon lever arm, determined from the literature (Krevolin et al., 2004). All grayscale ultrasound images gave regional attributes of dimension, position, coordinates and pixel grayscale values. In all compared frames the coordinates of the region of interest (ROI) in the frame were offset along the horizontal and vertical image planes and shifted by a pixel at a time to determine the degree of match.

Tracking algorithm

The algorithm used for tracking the ROI in the grayscale images has been described previously (Pearson et al., 2014). Briefly, a block matching method with normalised cross correlation (NCC) was utilised to determine similarity in the ROI between subsequent frames (images collected from video recorded at 25 Hz from start i.e. resting to maximum contraction force and split into subsequent frames for analysis). Here the ROI were arranged in two layers (Fig 1), with ROI 1 at 10mm from the patellar pole and the ROI 2 resting inter distance at 30mm. These ROI were then utilised to enable relative movement from the initial start, enabling strain to be determined for each layer region. As previously (Pearson et al., 2014), the ROI block size used was 15x15 pixels and the search window was 2x ROI (width) and 1x ROI size (height). For each frame search window comparison, the ROI were displaced by 1 pixel at a time from the original ROI start in the previous frame and compared using NCC. These data were then stored in a matrix for determination of best match based on the highest correlation values. The threshold level of correlation was set to ≥ 0.9. If this value was not reached or exceeded, the ROI was not shifted in the subsequent frame. If a correlation value above the threshold was detected, then the tracking resets the matching template and starts with the new updated position of the template ROI block in the next frame (Equation 1).

[Fig 1 near here]

$$\rho_{\mathrm{nm(k,l)}} = \frac{\sum_{i=-K}^{K} \sum_{j=-L}^{L} [F_1(n+i, m+j) F_2(n+k+i, m+l+j)]}{\left[\sum_{i=-K}^{K} \sum_{j=-L}^{L} [F_1(n+i, m+j)]^2 \right]^{1/2} \left[\sum_{i=-K}^{K} \sum_{j=-L}^{L} [F_2(n+k+i, m+l+j)]^2 \right]^{1/2}}$$

[1]

Where F1 is the start frame and F2 is the subsequent frame. The value (*n, m*) represents the image block centre, with the sum being over (*i ,j*), with (*k, l*) being the lateral and axial displacements respectively. The normalised cross correlation value (1 to -1, with 1 being the best match) is represented by *ρ nm(k. l).* The algorithm calculates this based on the two regions of interest (R_1 and R_2), (Fig 1), which are tracked from the first to the last frame, with the determined differences between the two ROI recorded up to maximum (100% MVC). Three trials were utilised to determine the means for each knee effort. Subsequent measures are anterior excursion and posterior excursion with knee angles ($90 - 30^0$). The total displacements of *x* and *y* over the frames is from determination of R_1 and R_2 measured from the first frame (f1) to the last frame (*fn*), with the resultant displacement as final frame - initial frame reference (Equation 2). Strain, therefore, is the displacement divided by the initial measure between R_1 and R_2.

$$\text{Disp} = \left(\sqrt{(x_{R2} - x_{R1})^2 + (y_{R2} - y_{R1})^2}\right)_{fn} - \left(\sqrt{(x_{R2} - x_{R1})^2 + (y_{R2} - y_{R1})^2}\right)_{f1}$$

[2]

Statistics

Intraclass coefficient correlations (ICC's) were utilised to determine the reliability of all measures. This was carried out by using repeated tracking on the regions during the ramped isometric contractions. Anterior vs. posterior strain comparisons by knee angle were made using repeated measures two-way ANOVA and Bonferroni *post hoc* pairwise tests. Paired t tests were utilised to compare strain ratio values at each knee angle and at fixed levels of force. The alpha level for all tests was set at 0.05. Statistical power was determined using G power (3.09, Franz Faul, Universitat, Kiel, Germany). For a power (1-β) of 0.8 and moderate effect size (0.4) a sample size of ten was determined.

RESULTS

Reliability

Pooled repeated displacement tracking analyses for the images of both layers gave an ICC of 0.88. The determined maximal mean percentage of hamstrings co-contraction (BF activity during maximal isometric knee extension), was seen to be 7.50 ± 0.30 %, which indicated a relatively low hamstring involvement during the maximal isometric agonist efforts.

Anterior vs. Posterior Strain

Tendon strain comparisons between the anterior and posterior regions showed significant differences pooled across the range of knee angles (anterior = 4.74 ± 1.95%; posterior = 3.57 ± 1.08%), ($p < 0.05$; $F(1, 9) = 89.81$; $\eta_p^2 = 0.90$; power = 1), with both regions declining as the knee joint angle decreased towards 30^0 (Figure 2), in line with the reductions in MVC force. Pooled strain at each knee angle showed significant differences with change in knee angle (90^0 = 6.38 ± 1.58%; 70^0 = 4.19 ± 0.73%, 50^0 = 3.34 ± 0.56%, 30^0 = 2.71 ± 0.47%), ($p < 0.05$; $F(1.56, 14.10) = 92.95$; $\eta_p^2 = 0.91$; power = 1).

The interaction of knee angle with region was also seen to be significant ($p < 0.05$; $F(1.83, 16.50) = 24.40$; $\eta_p^2 = 0.73$; power = 1), indicating differential strain by region at different knee angles . This was reflected in the strain ratios determined below.

[Figure 2 near here]

Regional strain ratios

The anterior region was found to exhibit significantly higher strain values than the posterior region at all angles of flexion 90^0 (7.76% ± 0.89 vs. 5.06% ± 0.76, $t(9) = 10.78$, $p < 0.001$, $r = 0.96$), 70^0 (4.77% ± 0.66 vs. 3.75% ± 0.38 $t(9) = 3.10$, $p = 0.013$, $r = 0.71$), and 50^0 (3.74% ± 0.37 vs. 2.90% ± 0.25, ($t(9) = 6.40$, $p < 0.001$, $r = 0.90$) apart from 30^0 (2.84% ± 0.53 vs. 2.57% ± 0.31, $p = 0.84$, $r = 0.54$) (Figure 2). Moreover, the magnitude of differences in strain between the anterior and posterior regions (anterior-to-posterior strain ratio) was greatest at 90^0 and lowest at 30^0, with the regional values being very similar at 30^0 knee flexion (Figure 3).

To account for differences in MVC force with change in knee angle, a standardised force level of 2117N (corresponding to the mean maximal force at 30^0 of knee flexion) was used to compare the anterior and posterior data sets. Once again, tendon strain was found to be higher in the anterior region compared to the posterior region at all knee joint angles, and the anterior-to-posterior strain ratio was greatest at 90^0 of flexion and lowest at 30^0 of flexion. (Figure 4). Comparisons of pooled anterior vs. posterior strain at the fixed level of force across the knee angles showed a significant difference (2.96 ± 0.43%; 2.35 ± 0.50%), ($t(3) = 4.27$, $p = 0.02$, $r = 0.92$).

[Figure 3 near here]

[Figure 4 near here]

DISCUSSION

To the authors knowledge this is the first study to report in vivo differential longitudinal strain in the patellar tendon with change in knee joint angle. The study examined the regional strain patterns of the patellar tendon during ramped isometric maximal contractions, over a range of knee joint angles. It was hypothesised that there would be differential intra-tendinous strain patterns, as well as variations in the anterior-to-posterior strain ratio with changes in knee joint angle. The findings of the present study have confirmed these hypotheses.

Tensile strain was found to be significantly greater ($p< 0.05$) in the anterior region than the corresponding posterior region at the knee joint angles of 90^0, 70^0 and 50^0. In addition, the anterior-to-posterior strain ratio (magnitude of differences in strain between anterior and posterior regions) was greatest at 90^0 (1.53) of knee flexion and lowest with the knee at 30^0 of flexion (1.10). These data add to previous observations showing intra-tendinous variations in loading patterns within the patellar tendon, and extend these observations by showing for the first time that the patellar tendon exhibits differential intra-tendinous strain characteristics specific to the knee joint angle at which loading is applied.

The mean maximal strain values of the patellar tendon reported here with the knee at 90^0 of flexion (7.76% ± 0.89) are in general agreement with previous work (Hansen et al., 2006; Malliaras et al., 2013; Pearson et al., 2014), who reported strains of 6.9 ± 0.6, 9-12% and 7.5-7.9% respectively . The higher strain reported by Malliaras and co workers may in part be due to subject characteristics i.e. they may have been initially untrained and so likely to have more compliant tendons. However strain with the knee at fixed flexion angles less than 90^0 as reported, here have not previously been examined in living specimens.

Previous research relating to the strain patterns of the anterior and posterior regions of the patellar tendon under load is diverse. Some authors have found the anterior region to display a significantly greater level of tensile strain relative to the posterior region (Almekinders et al., 2002; Pearson et al., 2014), whereas others have reported the posterior region to exhibit a greater magnitude of strain than the anterior region (Basso et al., 2002; Dillon et al., 2008). Despite such variability, which is perhaps, in part, attributable to inter-study differences in methodologies including the knee joint angle utilised , and also living vs. dead tissues (in vivo, vs. in vitro), the findings of the present study are in general agreement with those of Almekinders et al. (2002) and Pearson et al. (2014). Using real-time ultrasonography and dynamometry, Pearson and co workers reported a greater level of tensile strain in the anterior region compared to the posterior region at a knee joint angle of 90^0. Similarly, Almekinders and co-workers utilised cadaver knee specimens instrumented with strain gauges and reported tendon strain to increase at the anterior region and decrease at the posterior region as the knee was brought into flexion, although it is noteworthy that a compression of posterior tensile strain was not observed in the present study, perhaps due to the more complex interplay in living systems (antagonist/agonist muscle action and tissue characteristic differences) and the applied forces.

More specifically, in the present study, tendon strain values were found to be greater at the anterior region compared to the posterior region throughout the entire functional flexion range, with significantly higher values at all level of flexion apart from 30^0. These findings suggest that the posterior region of the patellar tendon is relatively stress shielded which could result in deterioration over time (Rufai et al., 1995; Vogel et al., 1993), leading to reduced mechanical strength and increased potential for injury. In line with this notion, previous studies on isolated patellar tendon fascicles have also shown decreased mechanical strength and stiffness at the posterior region compared to the anterior region (Hansen et al., 2010; Haraldsson et al., 2010), thus highlighting the lack of functional adaptation and weaker, more injury prone state of the posterior region.

Interestingly, the present study also found that the magnitude of differences in strain between the anterior and posterior regions was dependent on the knee joint angle at which loading was applied, with the greatest relative difference observed at 90^0 of flexion (1.53) and lowest at 30^0 of flexion (1.10). Such differential intra-tendinous strain patterns with the change in knee joint angle may be, in part, a factor of the alterations in the patellar lever arm arrangement with respect to the 'angle of pull' across the bony ends of the proximal and distal connections of the patellar tendon. Indeed, work by DeFrate et al. (2007) and Aalsberg et al. (2005), using MRI show changes in the patellar line of pull relative to the tibia with changes in knee flexion, suggesting a more 'posterior' pull on the tibia with increased knee flexion angle. It could be reasoned therefore that the alterations in the tendon alignment with respect to the bony anchors, and the patella with respect to the pull of the quadriceps muscle may result in changes on the intrinsic elements of the tendon resulting in differential regional strain as seen here. In addition, the notion that the turning moment of the patella increases with increased knee flexion is supported by the study of Ward et al. (2012). Here, fluoroscopic images showed an increase in the patella flexion angle with increased knee flexion such that the pull of the quadriceps against the patella becomes more angular to the patellar tendon (see figure 5 for illustration). One could thus envisage how this turning moment (action to lift the patella bone away from the patellar tendon as it glides around the intercondylar fossa of the femur), would generate more strain at the anterior aspect of the patellar tendon.

[Figure 5 near here]

The regional strain patterns observed in the present study are indeed indicative of shear stress within the patellar tendon during applied loading conditions, which is greatly implicated in the development and progression of patellar tendinopathy (Almekinders et al., 2002). The increased magnitude of tensile strain observed in the anterior region may suggest that it could ultimately reach a stage where it can no longer bear the functional demand and, as a result, becomes tendinopathic via ‘overuse’. On the other hand, the stress-shielded nature and lack of functional adaptation of the posterior region could also play an important role in the development of tendinopathy, simply owing to its weaker state (Almekinders et al., 2002; Orchard et al., 2004). Both these hypotheses can be seen to have some rationale here, i.e. unaccustomed loading of a 'relatively' weaker area of the tendon may result in some kind of insult to the structure, especially where patellar loading is carried out with the knee in flexion.

Interestingly, changes in transverse strain with loading observed by Wearing and co workers (2013) may be reflective of regional differences in longitudinal strain and it would be useful if measures of transverse strain could be more selective of layer or regions in order to enable relationships to longitudinal regional strain to be made.

Practical applications:

The findings here indicate that shear strain of the patellar tendon increases with increased knee flexion angle. Hence to enable a progressive return to play in sport where tendon injury has occurred it is recommended to begin loading with a reduced knee flexion angle and progress this in conjunction with load over time. Similarly to minimise effects of training induced damage, a range of angles should be utilised when preparing this structure for play, limiting maximal knee flexion wherever possible.

Conclusions:

In conclusion, we have shown that by varying the knee flexion angle we not only observe changes in absolute patellar tendon strain, but that the ratio between anterior and posterior tendon strain is also altered. The understanding that shear inducing difference between the surface and deep layers of the tendon occur differentially with knee angle may have implications for practices to help prevent tendon related injury and also aid targeted rehabilitation strategies.

The authors wish to acknowledge potential limitations with the current methodology, such that ROI identification and hence strain patterns during subsequent loading may be affected by out of plane movement. However using the current technology it is not possible to assess this effectively. We also acknowledge that the work presented here is in 2D and that future work would aim to extend this to a 3D scenario to detail further the characteristics of tendon strain under loading.

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Figure legends

Figure 1. Showing an example ultrasound image of the patellar tendon. Overlaid are the regions of interest seen as red squares (ROI), tracked with increasing force production. It can be seen that the ROI move laterally along the image with subsequent force production up until maximum for both anterior and posterior layers.

Figure 2. Mean values (± SD) of maximal strain values for all measured regions at each knee angle. (■) anterior region and (□) posterior region. * Significantly different $p< 0.05$, (anterior vs. posterior). ^ significantly different $p< 0.05$, (pooled strain).

Figure 3. Strain ratio determined at maximal strain for both anterior and posterior regions for each knee angle.

Figure 4. Strain ratio determined at a fixed force of 2117N (mean maximal force at knee angle of 30^0), at each knee angle.

Figure 5. Illustration showing potential turning effect of patella on tendon with knee flexed.