

Mechanical properties of the human achilles tendon

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Abstract

Objective. To determine whether the human Achilles tendon has higher material properties than other tendons and to test for strain rate sensitivity of the tendon.

Design. Mechanical testing of excised tendons.

Background. While the human Achilles tendon appears to experience higher in vivo stresses than other tendons, it is not known how the Achilles tendon's material properties compare with the properties of other tendons.

Methods. Modulus, failure stress, and failure strain were measured for excised human Achilles tendons loaded at strain rates of 1% s⁻¹ and 10% s⁻¹. Paired *t*-tests were used to examine strain rate effects, and average properties from grouped data were used to compare the Achilles tendon's properties with properties reported in the literature for other tendons.

Results. Failure stress and failure strain were higher at the faster strain rate, but no significant difference in modulus was observed. At the 1% s⁻¹ rate, the mean modulus and failure stress were 816 MPa (SD, 218) and 71 MPa (SD, 17), respectively. The failure strain was 12.8% (SD, 1.7) for the bone-tendon complex and 7.5% (SD, 1.1) for the tendon substance. At the 10% s⁻¹ rate, the mean modulus and failure stress were 822 MPa (SD, 211) and 86 MPa (SD, 24), respectively. The mean failure strain was 16.1% (SD, 3.6) for the bone-tendon complex and 9.9% (SD, 1.9) for the tendon substance. These properties fall within the range of properties reported in the literature for other tendons.

Conclusions. The material properties of the human Achilles tendon measured in this study are similar to the properties of other tendons reported in the literature despite higher stresses imposed on the Achilles tendon in vivo.

Relevance

The human Achilles tendon does not adapt to high in vivo stresses by developing correspondingly high material properties. This leaves the tendon at an increased risk of injury and may help to explain the high incidence of Achilles tendon injuries. Published by Elsevier Science Ltd.

Keywords: Achilles tendon; Mechanical properties; Failure; Adaptation

1. Introduction

The Achilles tendon is one of the most frequently injured tendons in the human body. It is the most frequently ruptured tendon [1,2] and, along with the patellar tendon, is one of the two tendons most frequently injured as a result of overuse [3]. The incidence of Achilles tendon injuries is rising, with ruptures occurring six times more frequently during the eight-year period ending in 1994 than during the previous eight years [4]. Achilles tendon injuries often cause severe, persistent pain and disability [5,6].

The high incidence of Achilles tendon injuries is related to the mechanical loading imposed on the tendon during physical activity. Achilles tendon ruptures most frequently occur in normally sedentary individuals engaged in intermittent strenuous physical activity [6,7]. Many overuse injuries are attributed to changes in activity such as abrupt increases in the duration or intensity of athletic training [3,6,8]. In both cases, injury appears to result from the tendon's inability to withstand the mechanical loading associated with increased physical activity.

It has been suggested that the human Achilles tendon experiences higher in vivo stresses than most other tendons. Ker et al. [9] used cross-sectional area ratios to estimate the in vivo stresses imposed on various tendons.

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They estimated that while most tendons experience peak stresses below 30 MPa, the human Achilles tendon experiences peak stresses around 67 MPa. Buckle transducer data also suggest that in vivo stresses are high in the human Achilles tendon. Komi et al. [10] reported peak stresses of 59 MPa for walking and 111 MPa for running in two subjects who had buckle transducers attached to their Achilles tendons.

Tendons have the ability to adapt to mechanical loading. Exercise can lead to increases in tendon modulus and strength [11,12] while immobilization or stress shielding can lead to decreases in these properties [13,14]. Following the principles of functional adaptation, we would expect the human Achilles tendon to adapt to high in vivo stresses by increasing its modulus and strength [15]. However, it is not clear whether the human Achilles tendon does indeed develop a higher modulus and strength than other tendons.

Material properties have been reported for many tendons. Modulus values are generally in the range of 500–1850 MPa [16–18]. Failure stresses are in the 50–125 MPa range ([19–21,16,18,22]). Nominal failure strains are 13–32% for bone-tendon-bone specimens [20,21] and 5–16% for the tendon midsubstance [16,18,22].

The material properties of tendons and ligaments depend, in part, on the rate at which the tendon or ligament is loaded. Most studies find higher failure stresses and failure strains at faster strain rates although changes in modulus are often not observed [23–27]. The rate-related increases in failure stress and failure strain are relatively modest. These properties increase 10–77% with 2–4 decades of increase in strain rate [24–27]. Although most studies have not found rate-related changes in modulus, two studies have noted modest increases in modulus with increased strain rate. Over four decades of strain rate, Hubbard and Soutas-Little [28] observed a 13% change in modulus for human palmaris longus and extensor hallucis longus tendons. Danto and Woo [17] found 31% and 94% changes in the moduli of rabbit Achilles and patellar tendons over four decades of strain rate.

In one of the few studies involving human Achilles tendon, Thermann et al. [29] reported mechanical properties for two displacement rates equivalent to strain rates of approximately 3% and 30% s⁻¹. These investigators did not find statistically significant differences in properties between the two rates. For the faster rate, they reported mean failure stresses of 41.4 MPa (SD, 10.3) and mean failure strains of 44.3% (SD, 15.3). For the slower rate, they reported mean failure stresses of 38.4 MPa (SD, 8.9) and mean failure strains of 49.2% (SD, 16.4). These failure stresses are lower and failure strains higher than the properties reported for other tendons. Thermann et al. did not report modulus values, but it can be inferred from their stiffness, length, and

cross-sectional area results that they obtained moduli much lower than those reported for other tendons.

Based on the results of Thermann et al., it would appear that the human Achilles tendon has a lower modulus and strength than other tendons and that its properties are not sensitive to strain rate. This runs contrary to our expectations that the Achilles tendon should adapt to high in vivo stresses by developing high modulus and strength and that it should exhibit strain rate sensitivity similar to other tendons. In this study, we try to determine whether the Achilles tendon has low modulus and strength, as suggested by Thermann et al., or high modulus and strength, as predicted by functional adaptation. We also test for strain rate sensitivity of the tendon's material properties.

2. Methods

Eleven pairs of fresh frozen Achilles tendons were procured from human donors aged 35–80 yr (mean 56.8, SD 13.3). The ages of all donors are listed in Table 1. Sources of the specimens included the International Institute for the Advancement of Medicine (Scranton, PA, USA), the Northern California Transplant Bank (San Rafael, CA, USA), and the Stanford University Anatomy Laboratory (Stanford, CA, USA). All specimens included the posterior half of the calcaneus and extended proximally to the mid-calf with intact calcaneal and muscle attachments. All soft tissue including the paratenon was removed from around the tendon except for the associated muscle, and the specimens were kept hydrated at all times. The tendons were screened for degenerative changes visually and through ultrasound examination, and no evidence of degeneration was observed. The calcanei were screened using dual-energy X-ray absorptiometry, and all specimens had a calcaneal bone mineral density above 0.5 g cm⁻². Specimens with a BMD below 0.5 g cm⁻² were excluded from this study.

Table 1
Specimen age and failure mode

Donor	Age	Failure mode (left)	Failure mode (right)
1	35	Midsubstance	Avulsion
2	44	Midsubstance	Midsubstance
3	47	Midsubstance	Midsubstance
4	52	Midsubstance	Midsubstance
5	55	Midsubstance	Midsubstance
6	55	Midsubstance	Midsubstance
7	56	Midsubstance	Midsubstance
8	57	Avulsion	Avulsion
9	65	Midsubstance	Midsubstance
10	79	Midsubstance	Avulsion
11	80	Midsubstance	Midsubstance

The tendon cross-sectional area was measured using ultrasound images. Previous studies have shown that this method of cross-sectional area measurement has good accuracy with average error less than 5% [36]. For each specimen, a transverse ultrasound image was recorded at each of three locations along the length of the tendon (2, 4, and 6 cm from the calcaneal insertion). NIH Image (US National Institutes of Health) was used to determine the tendon's cross-sectional area on each image. The minimum of the three area measurements (A_{\min}) was used in calculating stresses.

The calcaneus was embedded in polymethylmethacrylate (PMMA) and secured in an aluminum fixture (Fig. 1). The musculo-tendinous tissue was gripped in a liquid nitrogen-cooled freeze clamp [30] with a gage length of approximately 10 cm between the bottom of the freeze clamp and the top of the PMMA block. Six to 10 small beads (1 mm diameter) were glued at intervals (1–1.5 cm) along the length of the tendon to serve as visual markers.

Testing was performed at room temperature ($\sim 24^{\circ}\text{C}$) on a servo-hydraulic materials testing machine (MTS, Eden Prairie, MN, USA). Two strain rates were used. One specimen in each pair was tested at 1 mm s^{-1} , and the contralateral specimen was tested at 10 mm s^{-1} . These rates correspond with approximate strain rates of $1\% \text{ s}^{-1}$ and $10\% \text{ s}^{-1}$. The slower rate is considered quasi-static, while the faster rate is representative of physiologic activities such as walking. Each specimen was loaded to failure following 10 preconditioning cycles in which the specimen was stretched to 2% strain at 0.5 Hz.

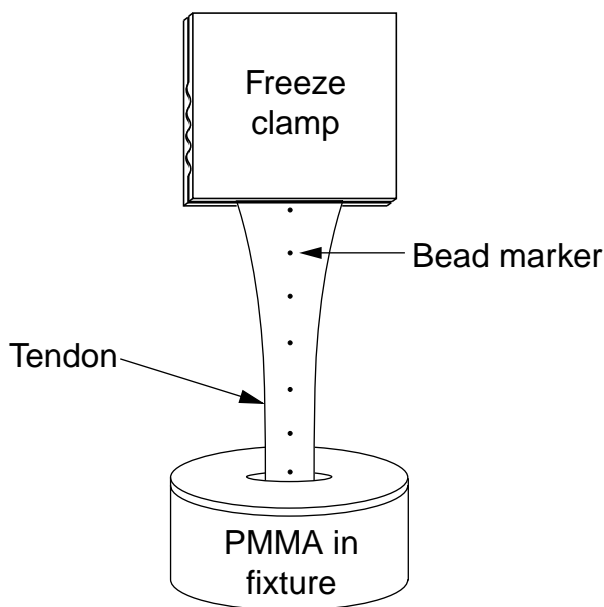


Fig. 1. Schematic of the testing setup. The calcaneus was embedded in PMMA and secured in a fixture. The musculo-tendinous tissue was gripped in a liquid nitrogen-cooled freeze clamp [30]. Bead markers 1 mm in diameter were attached to the tendon surface at 1–1.5 cm intervals.

During testing, a computer recorded forces (F) and crosshead displacements (d) while a CCD camera recorded the marker movements. The actual gage length of each specimen between the bottom of the freeze clamp and the top of the PMMA block (L_0) was determined from the video data. Stresses were defined as $\sigma = F/A_{\min}$, and strains were defined as $\varepsilon = d/L_0$. The failure load (F_{fail}) was defined as the maximum force recorded during each test. The displacement at failure (d_{fail}) was defined as the displacement at which the maximum load was achieved. The failure stress was calculated as $\sigma_{\text{fail}} = F_{\text{fail}}/A_{\min}$. The grip-to-grip failure strain for the entire bone-tendon complex was calculated as $\varepsilon_{\text{fail_complex}} = d_{\text{fail}}/L_0$. The modulus E was calculated as the slope of the stress-strain curve in the linear region beyond the initial toe region.

Tendon substance strains were determined from the video data. NIH Image (US National Institutes of Health) was used to measure the distance between the top and bottom markers in the first recorded frame (ℓ_0) and in the frame just prior to failure (ℓ_f). The nominal failure strain for the tendon substance was computed as $\varepsilon_{\text{fail_tendon}} = (\ell_f - \ell_0)/\ell_0$. Local failure strains were calculated in a similar manner using distances between adjacent markers in place of distances between the top and bottom markers.

Rate effects were studied using paired t -tests with the significance level set at $P < 0.05$. Since the modulus was expected not to change with strain rate, a two-tailed test was used for modulus. For all other properties, one-tailed tests were used since the properties were expected to be higher at the faster strain rate. Grouped data from each of the two displacement rates were used to determine average properties for the tendons. These average properties were then compared with the properties reported in the literature for other tendons including human patellar, semitendinosus, and gracilis tendons, rabbit Achilles and patellar tendons, and canine patellar tendons [16–22].

3. Results

Four specimens from three pairs failed by avulsion (Table 1). The three pairs were excluded from the paired analyses, and the four specimens were excluded from the grouped analyses. There was no clear relationship between failure mode and age. The avulsed specimens had a mean age of 57.0 yr (SD, 18.0). The remaining specimens all failed within the tendon substance and had a mean age of 56.8 yr (SD, 12.7). It was difficult to determine the exact location of initial failure, but it appeared on the video recordings that all non-avulsed specimens failed in the distal half of the tendon proximal to the bottom marker. The tendon has its smallest width in this region, and this region corresponds with the

Table 2
Mean (SD) mechanical properties of the human Achilles tendon from grouped data^a

	Rate		
	1 mm s ⁻¹ (n=9)	10 mm s ⁻¹ (n=9)	Combined (n=18)
Modulus (MPa)	816 (218)	822 (211)	819 (208)
Failure load (N)	4617 (1107)	5579 (1143)	5098 (1199)
Failure stress (MPa)	71 (17)	86 (24)	79 (22)
<i>Failure strain (%)</i>			
Bone-tendon complex	12.8 (1.7)	16.1 (3.6)	14.5 (3.2)
Tendon substance	7.5 (1.1)	9.9 (1.9)	8.8 (1.9)

^a The 1 mm s⁻¹ rate corresponds with a strain rate of approximately 1% s⁻¹. The 10 mm s⁻¹ rate corresponds with a strain rate of approximately 10% s⁻¹.

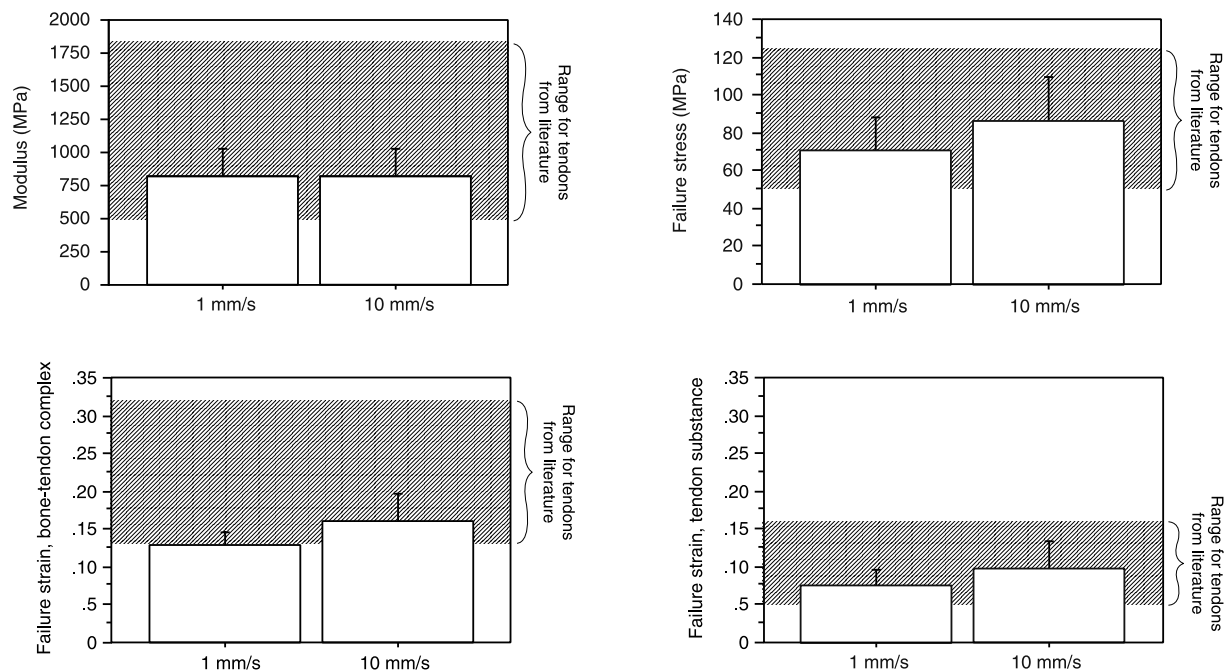


Fig. 2. Properties of human Achilles tendon from the current study (mean and SD) compared with properties of other tendons as reported in the literature [16–22]. The properties found in this study are in the lower end of the ranges reported in the literature.

location of most in vivo Achilles tendon ruptures [6]. No failures occurred at or near the freeze clamp.

Average properties were determined from grouped data at each rate for all specimens excluding those that failed by avulsion (Table 2). These properties are at the lower end of the ranges reported in the literature for other tendons (Fig. 2).

The paired *t*-tests indicated that there was no significant difference in modulus ($P = 0.61$) between the two rates, while failure load ($P = 0.01$), failure stress ($P = 0.03$), and failure strain ($P = 0.01$ for bone-tendon complex; $P = 0.04$ for tendon midsubstance) were all significantly higher at the faster rate (Fig. 3).

The local strains between adjacent markers ranged from 3% to 15% at failure. We observed no consistent patterns in the distribution of strains between regions,

but for each specimen at least one region had strains above 10% at failure. The greatest strains occurred below the bottom marker. Strains between the bottom marker and the top of the PMMA block were usually in excess of 20%. This accounted for the higher failure strains of the bone-tendon complex compared with the failure strains of the tendon substance.

4. Discussion

In this study, we measured the modulus, failure stress, and failure strain of human Achilles tendons at two strain rates. We found that the modulus did not differ between the two rates, while the failure stress and failure strain increased approximately 15% at the faster rate.

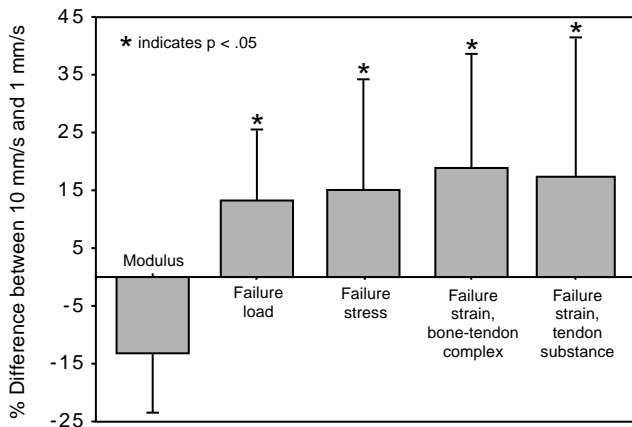


Fig. 3. Percent difference in properties between 10 mm s⁻¹ and 1 mm s⁻¹ displacement rates for pairs of specimens (mean and SD). The differences were significantly different from zero ($P < 0.05$) for all properties except modulus.

These rate effects are consistent with the rate effects reported in the literature for many other tendons and ligaments [23–27]. However, these effects differ from those reported by Lewis and Shaw for embalmed human Achilles tendons. Lewis and Shaw [31] found that the modulus was higher at 100% s⁻¹ than at 10% s⁻¹ while failure stress and failure strain did not differ between rates. Many factors may contribute to differences in the strain rate effects reported in various studies. Embalming may have altered the mechanical behavior of the tendons tested by Lewis and Shaw, and differences in the strain rates used may have lead to differences between their results and ours. Despite the differences between our results and those of Lewis and Shaw, however, the rate effects we observed are in agreement with rate effects reported for many other tendons and ligaments [23–27].

The faster strain rate (10% s⁻¹) we selected is characteristic of physiologic activities such as walking. The slower strain rate (1% s⁻¹) is considered quasi-static. Since many Achilles tendon injuries occur during loading at rates above 10% s⁻¹, it would be useful to obtain additional data at faster strain rates. While we were able to show statistically significant rate-related differences in failure stress and failure strain in this study, differences in modulus might also be observed with a wider range of rates.

This study considered two different measurements of failure strain. Strain of the entire bone-tendon complex was measured using grip-to-grip displacements, and strain of the tendon substance was measured using video analysis of marker movements. Consistent with previous studies [32], strains of the bone-tendon complex were higher than strains of the tendon substance. This is due in large part to large deformations near the calcaneal insertion below the bottom marker. Although approxi-

mately 20–30% of the total grip-to-grip deformation occurred below the bottom marker, none of our specimens failed in this high-deformation region. Currently, it is not known why deformations are higher in the insertional region or how the tissue is able to withstand the high strains in that region. Nevertheless, because tendon strains are higher near bony insertions, it is always advisable to differentiate between midsubstance strains and strains of a bone-tendon or bone-tendon-bone complex when considering failure strains.

One previous study measured mechanical properties of excised fresh human Achilles tendons [29]. While that study did not report modulus or failure strains of the tendon substance, it did report failure loads, failure stresses, and failure strains of the bone-tendon complex. The mean failure loads we measured were similar to those reported in that study. Thermann et al. found mean failure loads of 4635 N (SD, 923) and 4977 N (SD, 1168) for strain rates of approximately 3% s⁻¹ and 30% s⁻¹ compared with our results of 4617 N (SD, 1107) and 5579 N (SD, 1143) for strain rates of approximately 1% s⁻¹ and 10% s⁻¹.

In contrast, the failure stresses we measured were much higher than those reported by Thermann et al. This was due to higher cross-sectional area measurements in their study. They found a mean cross-sectional area of 127 mm² (SD, 39), resulting in failure stresses of 41 MPa (SD, 10) and 38 MPa (SD, 9) at their faster and slower strain rates, respectively. In contrast, we measured a mean cross-sectional area of 67 mm² (SD, 17) and mean failure stresses of 86 MPa (SD, 24) and 71 MPa (SD, 17) at our faster and slower rates. The differences in area measurement are likely due to differences in methods between the two studies. Thermann et al. made a single measurement and assumed a circular cross-sectional shape even though the cross-section is much better approximated by an ellipse. The assumption of a circular cross-sectional shape results in significant over estimation of the cross-sectional area.

For failure strain, we obtained much lower values than those reported by Thermann et al. They reported mean failure strains of 44.3% (SD, 15.3) and 49.2% (SE, 16.4) for their faster and slower rates, compared with our results of 16.1% (SD, 3.6) and 12.8% (SD, 1.7) for our two rates. Again, this difference is likely due in large part to differences in methods. Thermann et al. used specimens similar to ours, Achilles tendons with intact calcaneal insertions, but they used approximately half the length of tendon that we used. Since much of the deformation occurs near the calcaneal insertion, the shorter gage length leads to larger strain values for the bone-tendon complex. Other differences in methods likely account for the remaining difference between our results and those of Thermann et al.

Using cross-sectional area ratios, Ker et al. [9] estimated that the human Achilles tendon experiences peak

in vivo stresses around 67 MPa. Komi et al. [10] reported even higher stresses based on buckle transducer studies in which one subject produced a peak Achilles tendon stress of 59 MPa during walking and another subject produced a peak stress of 111 MPa during running. These stresses are quite high relative to the failure stresses we measured suggesting that the Achilles tendon is highly stressed during normal physical activities. High in vivo loading relative to the tendon's failure properties may contribute to the high incidence of Achilles tendon injuries.

The properties may also be affected by specimen age and testing protocol. While most studies have not found changes in tendon and ligament modulus, failure stress, and failure strain with age [21,22,26,28,33], a few studies have observed age-related changes in some of these properties. Lewis and Shaw [31] found statistically significant age-related decreases in the modulus and failure stress of embalmed human Achilles tendons. Noyes and Grood [34] observed decreases in human anterior cruciate ligament modulus and failure stress with age. Johnson et al. [18] found a decrease in the failure stress of human patellar tendons with age although they did not find age-related differences in modulus and failure strain. In this study, we observed trends towards decreased modulus and failure strain with age as well as a statistically significant age-related decrease in failure stress as determined by linear regression ($P = 0.02$).

Our tests were performed *ex vivo* using fresh frozen tendons. Effects due to freezing and storage are expected to be minimal, but the tests are still being performed on tissues outside the living body. Woo et al. [35] found that freezing and storage did not affect the load–deformation characteristics of rabbit medial collateral ligaments. Smith et al. [36] observed a small reduction in modulus but no change in ultimate tensile stress associated with freezing of porcine extensor tendons. Temperature effects are also expected to be minimal, and we would expect similar results at 37°C as we observed at room temperature. Although Woo et al. [37] observed changes in the viscoelastic behavior of canine medial collateral ligaments between 22°C and 37°C, Wang et al. [38] found that the moduli of wallaby and tiger tail tendons remained constant for temperatures from 20°C to 41°C and Hasberry and Percy [39] observed little change in the load–extension behavior of sheep interspinous ligaments between 19°C and 39°C. Our protocol did not include strain rates above 10% s⁻¹, and higher failure stresses and failure strains would be expected at faster strain rates. However, rate-related increases are generally only 10–75% over 2–4 decades of strain rate [23–27], and even a 75% increase would not place the Achilles tendon's properties above those reported for other tendons.

Our results suggest that the human Achilles tendon does not possess high material properties even though it

seems to experience high in vivo stresses. The properties we measured for the Achilles tendon are at the lower end of properties reported in the literature for other tendons. Even with possible increases in the properties due to increased strain rate or changes in testing procedures, the Achilles tendon properties would not be higher than the properties of other tendons with lower in vivo loading.

5. Conclusions

The human Achilles tendon exhibits strain rate sensitivity similar to other tendons, and it has material properties similar to other tendons despite higher in vivo loading. For reasons we do not yet understand, the tendon fails to adapt to high in vivo stresses by developing correspondingly high material properties. This failure to adapt leaves the tendon at increased risk of injury, especially when loading is suddenly increased by changes in physical activity. This may help to explain the relatively high incidence of Achilles tendon injuries.

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