Influence of aging on the in vivo properties of human patellar tendon


1Human Performance Laboratory, Ball State University, Muncie, Indiana; 2Institute of Sports Science and Clinical Biomechanics, University of Southern Denmark, Odense; and 3University of Copenhagen, Institute of Sports Medicine Copenhagen, Bispebjerg Hospital, Copenhagen, Denmark

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Carroll CC, Dickinson JM, Haus JM, Lee GA, Hollon CJ, Aagaard P, Magnusson SP, Trappe TA. Influence of aging on the in vivo properties of human patellar tendon. J Appl Physiol 105: 1907–1915, 2008. First published October 16, 2008; doi:10.1152/japplphysiol.00059.2008.—Tendons are important for optimal muscle force transfer to bone and play a key role in functional ability. Changes in tendon properties with aging could contribute to declines in physical function commonly associated with aging. We investigated the in vivo mechanical properties of the patellar tendon in 37 men and women (11 young [27 ± 1 yr] and 26 old [65 ± 1 yr]) using ultrasonography and magnetic resonance imaging (MRI). Patella displacement relative to the tibia was monitored with ultrasonography during ramped isometric contractions of the knee extensors, and MRI was used to determine tendon cross-sectional area (CSA) and signal intensity. At peak force, patellar tendon deformation, stress, and strain were 13 (P = 0.05), 19, and 12% less in old compared with young (P < 0.05). Additionally, deformation, stiffness, stress, CSA, and length were 18, 35, 41, 28, and 11% greater (P < 0.05), respectively, in men compared with women. After normalization of mechanical properties to a common force, no age differences were apparent; however, stress and strain were 26 and 22% higher, respectively, in women compared with men (P < 0.05). CSA and signal intensity decreased 12 and 24%, respectively, with aging (P < 0.05) in the midregion of the tendon. These data suggest that differences in patellar tendon in vivo mechanical properties with aging are more related to force output rather than an age effect. In contrast, the decrease in signal intensity indirectly suggests that the internal milieu of the tendon is altered with aging; however, the physiological and functional consequence of this finding requires further study.

ultrasonography; tendon injury; stress-strain relationship

IN CONTRAST TO skeletal muscle, studies in humans evaluating the influence of aging on in vivo tendon properties have been relatively limited (27, 32, 40). This lack of data is surprising given that tendons, once thought to be largely inert structures, have been shown to have specific mechanical properties (14) that can directly influence skeletal muscle performance and functional ability (3, 6, 40), and thus, any changes in tendon with aging may contribute to age-related declines in physical function (17). Additionally, observations of Achilles tendon injury rates (19, 29, 38) suggest that the incidence of tendon injuries may be increased in elderly individuals. Therefore, defining the influence of aging on tendon is important and will advance our understanding of the role of tendon in aging-associated tendon injury and declining physical function (17).

Over the past several years, various groups (5, 11, 14, 30, 45) have pioneered the use of ultrasonography for the evaluation of tendinous tissues in vivo and have provided a means to noninvasively determine the elastic properties of tendon in humans. Using this methodology, maximal human gastrocnemius tendon (40) and vastus lateralis tendon-aponeurosis (27) stiffness have been shown to decrease with aging, suggesting that tendon may become more compliant, which could influence muscle performance (6). These studies, however, measure elongation at the level of the gastrocnemius or vastus lateralis muscle, and the total elongation measured is likely influenced by other muscles involved in plantarflexion or knee extension, respectively. Additionally, free tendon and aponeurosis have different properties (34); thus this approach is inherently problematic when assessing free tendon adaptations because it is unlikely a measure of free tendon alone.

In addition to ultrasound-based studies, several in vitro studies have reported reductions in tendon ultimate tensile strength (20, 57, 59) and changes in tendon ultrastructure (8, 58) with aging. Yet, to our knowledge, no data exist describing the influence of aging on the in vivo mechanical properties of the patellar tendon in humans, an important tendon for performance of normal locomotion and activities of daily living. Therefore, we chose to utilize ultrasonography and MRI to determine 1) if the in vivo mechanical properties of the patellar tendon are altered with aging, 2) if patellar tendon size is decreased with aging, and 3) if aging alters MRI-determined signal intensity. On the basis of previous findings in humans (27, 40), we hypothesized that patellar tendon stiffness, modulus, strain, and stress would be lower in older individuals. More recent evidence (41) also suggests that patellar tendon mechanical properties are sex specific; therefore, we recruited both men and women to participate in this investigation. Given the absence of data available in humans describing tendon adaptations to aging, the findings from this study will contribute significant new knowledge to the areas of aging and human tendon physiology and provide better guidance for patient care and injury prevention.

METHODS

Subjects. Eleven young (6 men and 5 women) and 26 older (16 men and 10 women) individuals were recruited to participate in this investigation (Table 1). The Institutional Review Board of Ball State University and Ball Memorial Hospital approved this study, and informed consent was obtained from all subjects before participation. All subjects were sedentary to moderately active, nonsmokers, and apparently healthy as determined from a detailed medical history questionnaire and standard blood and urine chemistries. Additionally, the costs of publication of this article were defrayed in part by the payment of page charges. The article must therefore be hereby marked “advertisement” in accordance with 18 U.S.C. Section 1734 solely to indicate this fact.

Address for reprint requests and other correspondence: T. A. Trappe, Human Performance Laboratory, Ball State Univ., Muncie, IN 47306 (e-mail: trtrappe@bsu.edu).
all older individuals were given a medical exam and completed a resting and exercise electrocardiogram. Any individuals with known knee or tendon pathologies were excluded.

Muscle strength assessment: one-repetition maximum. Quadriceps strength was evaluated by assessing each individual’s bilateral one-repetition maximum (1RM) on a knee extension device (Cybex Eagle, Medway, MA). After 5 min of light cycling and two warm-up sets of five repetitions, several 1RM attempts were performed each separated by a 2-min rest period. The maximum weight lifted through a full range of motion was recorded as the subject’s 1RM.

MRI: quadriceps femoris. After lying supine for 1 h (2), axial images of the quadriceps femoris (55) were obtained [repetition time (TR)/echo time (TE) = 2,000/9; field of view (FOV) = 48 cm, 512 × 512 matrix; slice thickness = 8 mm; spacing = 0 mm] for the determination of muscle size. Muscle image files were transferred to a personal computer (iMac G5) and analyzed with NIH Image software (ImageJ, version 1.34) using manual planimetry, as done previously in our laboratory (54, 55). The same investigator performed all analyses.

MRI: patellar tendon. Immediately following the muscle scan, sagittal (Fig. 1A) and axial (Fig. 1B) images of the patellar tendon were obtained. While continuing to lie supine, the individual’s right knee was placed in an extremity coil (GE 1.5 T, Quadrature Lower Extremity Coil 472GE-64, Invivo, Pewaukee, WI). Leg and joint images of the quadriceps femoris (55) were obtained [repetition time (TR)/echo time (TE) = 48 cm, 512 × 512 matrix; slice thickness = 4 mm; spacing = 0 mm] for the determination of tendon signal intensity. Sagittal images were obtained [TR/TE = 400/14, spin echo; echo train length (ETL) = 0; number of excitations (NEX) = 2; percent phase field of view (PPFOV) = 100; FOV = 16 × 16 cm, 256 × 256 matrix; slice thickness = 4 mm; spacing = 0 mm] beginning on the lateral most portion of the lateral condyle of the tibia, then moving medially (based on a coronal scout scan). Axial images of the patellar tendon were obtained [TR/TE = 550/12.4, fast spin echo train; ETL = 3; NEX = 3; PPFOV = 100; FOV = 16 × 16 cm, 256 × 256 matrix; slice thickness = 4 mm; spacing = 0 mm] beginning 8 mm (2 slices) proximal of the distal pole of the patella and proceeding distally (based on a sagittal scout scan). Tendon image files were then transferred to a personal computer (iMac G5) for analysis. For tendon length, sagittal images were reviewed with OsiriX (software version 2.7.5), and only slices with complete tendon from the distal pole of the patella to tibial insertion were chosen for analysis. ImageJ (version 1.34) was then used to determine tendon length via manual planimetry. The shortest measured distance from the most distal part of the patella to the first insertion point on the tibia along the posterior portion of the tendon was recorded as tendon length (22).

Axial scans were also reviewed using OsiriX, and then ImageJ was used to manually circumscribe the patellar tendon. Specifically, tendon CSA and signal intensity [mean gray value (MGV)] were determined from an average of all slices beginning with the first slice not containing patella, then proceeding distally until just before insertion of the patellar tendon into the tibia (i.e., no indication of bone contacting the tendon). Tendon signal intensity was normalized to the 1% CuSO4 (tendon signal intensity/CuSO4 signal intensity). To determine region-specific CSA and signal intensity, the first slice not containing patella was considered the proximal tendon, and the slice just before insertion of the patellar tendon into the tibia was considered the distal tendon. The midregion was taken as the slice halfway between the proximal and distal slices. If a subject had an even number of slices, the values from the two slices nearest the midregion were averaged. An average of five traces per slice were used for the calculation of length, CSA, and signal intensity, and the same investigator completed all measurements. The coefficient of variation for trace-retrace was 1.1, 2.9, and 3.8% for tendon length, CSA, and signal intensity, respectively.

Tendon mechanical properties. Patellar tendon mechanical properties were assessed as previously described (14, 22) with modifications. Specifically, subjects were firmly strapped into a rigid aluminum chair with the knee joint at 90° of flexion. A slightly padded cuff was attached to their lower leg ~3 cm proximal to the medial malleolus. All participants completed three sessions, one familiarization and two testing. For each session, subjects performed four to seven “ramped” 10-s isometric quadriceps contractions to maximal effort with 90-s rest between contractions. The goal of each session was to obtain four attempts with linear increases in force; however, if during testing it was felt that there was an error during the contraction, e.g., reaching peak force too early, excessive subject movement, or issues with probe contact, an additional ramped contraction was performed after a 90-s rest period. During each contraction force was recorded (60-Hz sampling rate) via a strain-gauge load cell (Omega Dynac, LC101-500) interfaced with a personal computer (Gateway E-4200). During ramped contractions an ultrasound probe (7.5 MHz, 70-mm B-mode linear array, Sonoline Sierra, Siemens, Erlangen, Germany) was mounted on the skin overlying the patellar tendon (Fig. 2) to monitor displacement of the patella and tibia during the ramped isometric effort (14). Video feed from the ultrasound unit was captured on a personal computer using frame grabber software (Matrox Inspector, Matrox Electronic Systems, Dorval, Quebec, Canada). To permit later synchronization of the ultrasound video and force output, a custom device was built that provided a visual signal on each ultrasound video allowing the user to select the specific frame of video at which force collection commenced.

Placement of the patella relative to the tibia was determined with custom-designed software (34), and deformation was defined as the change in distance between the tibia and patella, as previously described (14). Tendon force was calculated by dividing the total knee extension moment by the internal moment arm, which was estimated from femur length (14, 58). For each ramped contraction, deformation was determined three times and the highest observed deformation was used for analysis. After determining the deformation for each ramped contraction, the two contractions with the highest and lowest deformation were excluded. Force and deformation data from the remaining attempts were analyzed to a greatest common force and averaged.

### Table 1. Subject characteristics

<table>
<thead>
<tr>
<th>Group</th>
<th>Age (yr)</th>
<th>Height (cm)</th>
<th>Body Weight (kg)</th>
<th>BMI (kg/m²)</th>
<th>1RM (kg)</th>
<th>1RM/Body Weight</th>
<th>Quadriceps CSA (mm²)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Young (n = 11)</td>
<td>27±1</td>
<td>171±3</td>
<td>71.9±4.7</td>
<td>25±1</td>
<td>93.9±9*</td>
<td>1.3±0.1*</td>
<td>6,814±488*</td>
</tr>
<tr>
<td>Old (n = 26)</td>
<td>65±1*</td>
<td>171±2</td>
<td>84.0±2.5*</td>
<td>29±1</td>
<td>68.5</td>
<td>0.8±0.1</td>
<td>5,684±460</td>
</tr>
<tr>
<td>YM (n = 6)</td>
<td>27±2</td>
<td>176±3</td>
<td>82.9±4.9</td>
<td>27±2</td>
<td>113.5</td>
<td>1.4±0.1</td>
<td>8,030±396</td>
</tr>
<tr>
<td>OM (n = 16)</td>
<td>65±1</td>
<td>179±3</td>
<td>89.4±2.6</td>
<td>28±1</td>
<td>85.3</td>
<td>1.0±0.1</td>
<td>6,713±223</td>
</tr>
<tr>
<td>YW (n = 5)</td>
<td>29±1</td>
<td>164±3</td>
<td>58.8±2.8</td>
<td>32±1</td>
<td>62.5</td>
<td>1.1±0.1</td>
<td>5,355±316</td>
</tr>
<tr>
<td>OW (n = 10)</td>
<td>66±2</td>
<td>160±4</td>
<td>75.2±3.4</td>
<td>30±1</td>
<td>40±2</td>
<td>0.5±0.1</td>
<td>4,038±156</td>
</tr>
</tbody>
</table>

Data are expressed as means ± SE; n = no of subjects. 1RM, one-repetition maximum; BMI, body mass index; CSA, cross-sectional area; YM, young men; OM, old men; YW, young women; OW, old women. *P < 0.05, Young vs. Old. †P < 0.05, YW different from OM and OW.
The familiarization session was not used in the data analysis; thus the results of the second and third testing sessions were averaged. The same individual performed all mechanical analyses. The average coefficient of variation between the two included trials from each session (within day) was 4.4, 7.8, 7.8, 0.8, and 4.4% for tendon deformation, stiffness, modulus, stress, and strain, respectively. Average between-day coefficient of variation was 8.7, 7.0, 7.0, 0.6, and 8.7% for tendon deformation, stiffness, modulus, stress, and strain, respectively. The average values obtained between testing sessions 2 and 3 were not different ($P > 0.05$) for tendon deformation (2.6 ± 0.2 vs. 2.7 ± 0.2 mm), stiffness (2.383 ± 0.4 vs. 2.341 ± 0.4 N/mm), modulus (0.93 ± 0.33 vs. 0.91 ± 0.31 GPa), stress (23.5 ± 0.4 vs. 23.5 ± 0.4 MPa), and strain (5.5 ± 0.2 vs. 5.7 ± 0.2%). These values are similar to a previous evaluation of patellar tendon in vivo mechanical properties in young individuals (14).

Normalized force comparison. For further comparison of mechanical properties, tendon deformation, stiffness, modulus, stress, and strain were determined for each individual at a common force. We initially chose to compare all groups at the lowest common mean force, i.e., the average tendon force of older women (1,650 N). However, given the large force differences between men and women, we also felt that comparing men and women separately was warranted. In this case, mechanical properties for young and old men were calculated at the mean tendon force of old men (3,014 N), and mechanical properties of young and old women were calculated at the mean force of old women (1,650 N).

Statistics. A one-way ANOVA was used to compare tendon CSA and MGV at proximal, mid, and distal tendon. All other data were compared using two-way (age and sex) ANOVA. Tukey’s honestly significant difference test was used to explore differences when a significant interaction was detected. Values were considered significant at an $α$-level of $P < 0.05$. All data were expressed as means ± SE.

RESULTS

No age and sex interactions were detected for any of the patellar tendon properties measured; therefore, data were collapsed and presented as young vs. old (age) and men vs. women (sex). Individual group data are also presented for general discussion.

MRI-determined tendon properties. When all slices along the tendon were averaged, absolute patellar tendon CSA was not affected by aging (Table 2). In contrast, after normalization to body weight, tendon CSA was lower in the old individuals ($P < 0.05$, Table 2). Additionally, the CSA of the tendon midregion was 12% ($P < 0.05$) lower in older individuals (Fig. 3B). Averaged tendon CSA along the tendon was 28% greater ($P < 0.05$) in men compared with women (Table 2), and this difference was consistent along the tendon (Fig. 3C). However, this sex effect was eliminated after correction for body weight (Table 2). Tendon CSA was directly correlated ($P < 0.05$) to 1RM strength ($P < 0.05$, $r = 0.68$), quadri-
with aging. These differences, however, appear to be more related to force output rather than an age effect. Although tendon CSA and MRI signal intensity decreased with aging, aging does not appear to alter the slope of the force-deformation (stiffness) and stress-strain (modulus) relationships of the patellar tendon.

Our findings suggest that tendon elastic properties, at least of the patellar tendon, may not explain age-related declines in quadriceps muscle strength or functional performance (17).

decreased 18% with aging (P < 0.05, Table 2), but there was a trend for signal intensity to be greater in women at the proximal tendon (P = 0.054, Fig. 3C).

Comparison at maximal tendon force. Young individuals were 37% stronger (P < 0.05, Table 1) than old subjects. At maximal levels of tendon force (Table 3), older individuals had 13% less patellar tendon deformation (P = 0.05; Fig. 4A) and lower peak stress (19%) and strain (12%, Fig. 5A) (P < 0.05) compared with young. Aging did not alter maximal tendon stiffness (Fig. 4A) or elastic modulus (Fig. 5A, Table 3) of the patellar tendon.

Men (1RM = 93 kg) were 100% stronger (P < 0.05) than women (1RM = 46 kg). At maximal levels of tendon force, patellar tendon deformation (18%), stiffness (35%, Fig. 4C), and stress (41%, Fig. 5C) were greater in men compared with women (P < 0.05, Table 3). There was no difference between men and women in either maximal patellar tendon modulus or strain (Table 3, Fig. 5C).

Normalized force comparisons. When calculated at relative levels of tendon force, no age differences in patellar tendon mechanical properties were apparent (Table 4; Figs. 4B and 5B). However, patellar tendon stress and strain were 26 and 22% higher (Table 4, Fig. 5D), respectively, in women compared with men regardless of age (P < 0.05).

DISCUSSION

Tendon properties and age. To our knowledge, there are no data describing the influence of aging on in vivo patellar tendon properties in humans. Therefore, we evaluated the in vivo mechanical properties of the patellar tendon in young and old men and women. Consistent with mechanical data from other tendon structures (27, 32, 40), we demonstrate that tendon stress and strain at peak force output are reduced with aging. These differences, however, appear to be more

Table 2. MRI-determined patellar tendon properties

<table>
<thead>
<tr>
<th></th>
<th>Tendon CSA, mm²</th>
<th>Tendon CSA/Body Weight, mm²/kg</th>
<th>Signal Intensity, MGV</th>
<th>Tendon Length, mm</th>
</tr>
</thead>
<tbody>
<tr>
<td>Young (n = 11)</td>
<td>127.5 ± 8.7</td>
<td>1.8 ± 0.2*</td>
<td>22.3 ± 1.0*</td>
<td>46.4 ± 2.1</td>
</tr>
<tr>
<td>Old (n = 26)</td>
<td>117.5 ± 4.7</td>
<td>1.4 ± 0.2</td>
<td>18.3 ± 0.9</td>
<td>45.8 ± 1.2</td>
</tr>
<tr>
<td>Men (n = 22)</td>
<td>131.9 ± 5.1†</td>
<td>1.5 ± 0.2</td>
<td>18.4 ± 0.9</td>
<td>48.0 ± 1.2†</td>
</tr>
<tr>
<td>Women (n = 15)</td>
<td>103.1 ± 4.6</td>
<td>1.5 ± 0.2</td>
<td>21.1 ± 1.2</td>
<td>43.1 ± 1.5</td>
</tr>
<tr>
<td>YM (n = 6)</td>
<td>145.4 ± 9.0</td>
<td>1.8 ± 0.2</td>
<td>22.3 ± 1.6</td>
<td>50.7 ± 1.9</td>
</tr>
<tr>
<td>OM (n = 16)</td>
<td>126.8 ± 5.8</td>
<td>1.4 ± 0.2</td>
<td>17.0 ± 0.9</td>
<td>47.0 ± 1.5</td>
</tr>
<tr>
<td>YW (n = 5)</td>
<td>106.1 ± 9.3</td>
<td>1.8 ± 0.1</td>
<td>22.3 ± 1.2</td>
<td>41.3 ± 2.3</td>
</tr>
<tr>
<td>OW (n = 10)</td>
<td>101.6 ± 5.4</td>
<td>1.4 ± 0.1</td>
<td>20.5 ± 1.8</td>
<td>44.0 ± 2.0</td>
</tr>
</tbody>
</table>

Data are expressed as means ± SE; n = no. of subjects. MGV, mean gray value. *P < 0.05, Young vs. Old. †P < 0.05, men vs. women.
This lack of change is surprising given 1) the large decrease in muscle mass and strength associated with aging (also seen in this study); 2) the acute and chronic influence of skeletal muscle loading (24, 26–28, 37, 44, 46) and unloading (7, 23, 31, 43) on tendon in humans; and 3) several in vitro studies in humans (10, 48) and animals (18, 56), which have demonstrated changes in tendon cellular structure with aging. However, it is plausible that the reduced loading of the patellar tendon with age may result in structural alterations (e.g., changes in collagen cross-linking or content) within the tendon, influencing the integrity of the tendon, which were not detected with the ultrasound measures. Consistent with this argument, patellar tendon collagen content of animals has been shown to be altered with age (18, 56). Additionally, Johnson et al. (20) have reported that in vitro patellar tendon ultimate tensile strength is reduced with aging (young, 64.7 MPa; old, 53.6 MPa). These measurements, however, were taken at the failure point of the tendon and likely do not reflect the actual conditions (e.g., tensile forces) reached in older individuals during daily activities (21.6 ± 2.4 MPa in our older subjects at peak force). Although speculative, the lower force-generating capacity of sedentary older individuals may aid in minimizing the development of tendinopathies in these individuals by providing a buffer to excessive

Table 3. Tendon mechanical properties at peak force

<table>
<thead>
<tr>
<th></th>
<th>Deformation, mm</th>
<th>Stiffness, N/mm</th>
<th>Modulus, GPa</th>
<th>Stress, MPa</th>
<th>Strain, %</th>
</tr>
</thead>
<tbody>
<tr>
<td>Young (n = 11)</td>
<td>2.8±0.2‡</td>
<td>2,552±218</td>
<td>0.9±0.1</td>
<td>26.6±2.4*</td>
<td>6.0±0.2*</td>
</tr>
<tr>
<td>Old (n = 26)</td>
<td>2.4±0.1</td>
<td>2,411±170</td>
<td>0.9±0.1</td>
<td>21.6±1.5</td>
<td>5.3±0.2</td>
</tr>
<tr>
<td>Men (n = 22)</td>
<td>2.7±0.1†</td>
<td>2,744±178†</td>
<td>1.0±0.1</td>
<td>26.2±1.7†</td>
<td>5.6±0.2</td>
</tr>
<tr>
<td>Women (n = 15)</td>
<td>2.3±0.1</td>
<td>2,026±153</td>
<td>0.9±0.1</td>
<td>18.6±1.4</td>
<td>5.3±0.3</td>
</tr>
<tr>
<td>YM (n = 6)</td>
<td>3.1±0.2</td>
<td>2,924±332</td>
<td>1.0±0.1</td>
<td>30.3±3.6</td>
<td>6.0±0.2</td>
</tr>
<tr>
<td>OM (n = 16)</td>
<td>2.6±0.2</td>
<td>2,676±215</td>
<td>1.0±0.1</td>
<td>24.6±1.9</td>
<td>5.4±0.3</td>
</tr>
<tr>
<td>YW (n = 5)</td>
<td>2.4±0.2</td>
<td>2,106±64</td>
<td>0.8±0.1</td>
<td>22.1±1.7</td>
<td>6.0±0.4</td>
</tr>
<tr>
<td>OW (n = 10)</td>
<td>2.2±0.1</td>
<td>1,986±231</td>
<td>0.9±0.1</td>
<td>16.8±1.8</td>
<td>5.0±0.3</td>
</tr>
</tbody>
</table>

Data are expressed as means ± SE. *P < 0.05, Young vs. Old. †P < 0.05, men vs. women. ‡P < 0.05, Young vs. Old.

Fig. 4. Mean tendon force (y-axes; in N) vs. deformation curves. Data fitted with 2nd-order polynomial regression. Slope of the line equals tendon stiffness (N/mm). A: young vs. old at maximal tendon force. *P < 0.05, force and deformation, young > old; for deformation P = 0.05, young > old. B: young vs. old at normalized tendon force. C: men vs. women at maximal tendon force. *P < 0.05, force, deformation, and stiffness, men > women. D: men vs. women at normalized tendon force.
tendon strain. This concept, however, may only be applicable to the patellar tendon, as the incidence of Achilles tendon ruptures appears to be elevated in elderly individuals (19, 38).

Interestingly, although the average CSA along the tendon was not different with aging, we did observe regional tendon atrophy (Fig. 3B), the implications of which require further study. However, these data do suggest that stress may vary along the length of the tendon, potentially predisposing the midregion of the tendon to injury. Interestingly, after adjusting for body weight, average CSA along the tendon was lower in our older individuals. We can speculate that the greater body weight (Table 1) of the older individuals may act as a loading stimulus, minimizing any age-related decrease in absolute tendon CSA.

In addition to our measure of patellar tendon CSA, the MRI allows for the determination of signal intensity, which reflects the gray pixel density of a given tissue. Signal intensity has previously been shown to increase with tendinosis (50) and may indicate changes in extracellular matrix, tissue hydration,

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**Table 4. Patellar tendon mechanical properties: force-normalized comparison**

<table>
<thead>
<tr>
<th>Deformation, mm</th>
<th>Stiffness, N/mm</th>
<th>Modulus, GPa</th>
<th>Stress, MPa</th>
<th>Strain, %</th>
<th>% of Maximal Force</th>
</tr>
</thead>
<tbody>
<tr>
<td>All subjects at ~1,650 N</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Young (n = 11)</td>
<td>2.0±0.1</td>
<td>1,818±83</td>
<td>0.7±0.1</td>
<td>13.6±1.0</td>
<td>4.5±0.3</td>
</tr>
<tr>
<td>Old (n = 26)</td>
<td>2.1±0.1</td>
<td>1,929±100</td>
<td>0.8±0.1</td>
<td>14.8±0.6</td>
<td>4.6±0.2</td>
</tr>
<tr>
<td>Men (n = 22)</td>
<td>2.0±0.1</td>
<td>1,922±97</td>
<td>0.7±0.1</td>
<td>13.1±0.5*</td>
<td>4.2±0.2*</td>
</tr>
<tr>
<td>Women (n = 15)</td>
<td>2.2±0.1</td>
<td>1,858±142</td>
<td>0.8±0.1</td>
<td>16.5±0.8</td>
<td>5.1±0.2</td>
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<tr>
<td>Men at ~3,014 N</td>
<td></td>
<td></td>
<td></td>
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<td></td>
</tr>
<tr>
<td>YM (n = 6)</td>
<td>2.6±0.2</td>
<td>2,472±203</td>
<td>0.9±0.1</td>
<td>21.3±1.2</td>
<td>5.2±0.3</td>
</tr>
<tr>
<td>OM (n = 16)</td>
<td>2.6±0.2</td>
<td>2,656±194</td>
<td>1.0±0.1</td>
<td>24.7±1.2</td>
<td>5.5±0.3</td>
</tr>
<tr>
<td>Women at ~1,650 N</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>YW (n = 5)</td>
<td>2.1±0.1</td>
<td>1,772±63</td>
<td>0.7±0.1</td>
<td>16.0±1.5</td>
<td>5.2±0.4</td>
</tr>
<tr>
<td>OW (n = 10)</td>
<td>2.2±0.1</td>
<td>1,902±214</td>
<td>0.8±0.1</td>
<td>16.7±0.9</td>
<td>5.1±0.3</td>
</tr>
</tbody>
</table>

Data are expressed as means ± SE; n = no. of subjects. *P < 0.05, men vs. women.
collagen fiber structure, or proteoglycan and glycosaminoglycan content (9, 12, 51, 62). Interestingly, we observed a 18\% decline \((P < 0.05)\) in MRI signal intensity of the patellar tendon (Table 2) with aging. This effect was also region specific (Fig. 3B). Several investigations (49, 52, 53) have shown tendon MRI signal to be altered with tendinosis or tissue injury and change with healing or chronic exercise (50), highlighting the potential physiological relevance of the signal intensity measurement. However, further biochemical studies of the tendon are needed to determine the physiological significance of the signal intensity measurement.

The region specificity of MRI signal intensity is consistent with previous studies of the patellar tendon, which have found regional difference in tendon CSA (22, 35, 60), mechanical properties (15), and collagen and proteoglycan mRNA expression (47). Interestingly, the decrease in signal intensity approximately mirrors the change in CSA along the tendon. Although speculative, the tendon may alter its properties at points along the tendon to compensate for lower CSA to prevent excessive tendon strain.

**Tendon properties and sex.** Similar to previous studies (25, 41, 60), we found the in vivo properties of tendon to be different between men and women. Tendon stiffness and stress at maximal force were greater in men, which is consistent with previous reports (25, 41, 60). However, after normalization to a common force, the difference in stiffness was no longer observed. As with other studies of the patellar tendon, absolute tendon CSA (41, 60) and length (16) were greater in men compared with women. The sex difference in tendon CSA appears to be related to the greater body weight of men, as the sex difference in tendon CSA was eliminated after correction for body weight (Table 2). Therefore, the greater tendon CSA in men may be a body/muscle size effect rather than a true sex effect, which is supported by the correlations between body weight/muscle size and tendon CSA. The lack of difference in modulus (material stiffness) and MRI signal intensity would also suggest that the material properties of the tendon are similar between men and women. Therefore, sex differences in tendon mechanical properties appear to be more related to body size and muscle force output rather than an intrinsic difference in tendon. However, studies of isolated patellar tendon have shown that the modulus and ultimate stress of collagen fascicles are greater in men compared with fascicles from women (33). After normalization to a common force, patellar tendon stress and strain were greater in females compared with males, which may influence the risk of tendon injury. Based on our findings and those of others (33, 36, 60), further studies evaluating tendon properties in females are warranted.

**Limitations.** Although our measurement of tendon mechanical properties provides a unique way to noninvasively study the human tendon in vivo, there are some limitations. The MRI and knee coil configuration utilized in this study do not allow for the knee joint and patellar tendon to be evaluated at 90° of flexion, i.e., the angle at which tendon mechanical testing is performed. Although this may prevent the determination of absolute mechanical parameters, all subjects are evaluated in the same manner; thus between-group comparisons should not be compromised. The inherent variability of the ultrasound technique may have prevented us from detecting more subtle differences in tendon mechanical properties, especially given our MRI findings. This combined with the smaller sample size in the young groups may have limited our ability to detect differences between groups.

**Summary.** We demonstrate that although patellar tendon mechanical properties at maximal force are altered with aging, these differences are more related to force output rather than an age effect. Even with the large decrease in muscle mass and strength with aging, we demonstrate that the force-deformation (stiffness) and stress-strain (modulus) relationships are not altered, at least until the eighth decade of life, in healthy individuals. The lower stress and strain at maximal force in the old, due to a lower force output, suggests older individuals may have a lower risk of traumatic patellar tendon injury. However, one must consider the regional differences in tendon CSA and signal intensity, the latter of which may reflect structural alterations in the tendon, that were not detected by the ultrasound measures. Given the decrease in signal intensity with aging, we feel that additional studies are warranted to determine the physiological relevance of this measure.

In addition to our aging findings and in concurrence with previous studies (41, 60), in vivo patellar tendon mechanical properties are different in males compared with females. These differences appear to be largely related to anatomic difference (tendon size) in the patellar tendon between men and women and may have implications for the greater rate of tendon injury in physically active females (1, 4, 13, 21).

**ACKNOWLEDGMENTS**

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**REFERENCES**

AGE AND PATELLAR TENDON PROPERTIES IN HUMANS


