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Apparent Transverse Compressive Material Properties of the Digital Flexor Tendons and the Median Nerve in the Carpal Tunnel

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Abstract

Carpal tunnel syndrome is a frequently encountered peripheral nerve disorder caused by mechanical insult to the median nerve, which may in part be a result of impingement by the adjacent digital flexor tendons. Realistic finite element (FE) analysis to determine contact stresses between the flexor tendons and median nerve depends upon the use of physiologically accurate material properties. To assess the transverse compressive properties of the digital flexor tendons and median nerve, these tissues from ten cadaveric forearm specimens were compressed transversely while under axial load. The experimental compression data were used in conjunction with an FE-based optimization routine to determine apparent hyperelastic coefficients (μ and α) for a first-order Ogden material property definition. The mean coefficient pairs were $\mu=35.3\text{kPa}$, $\alpha=8.5$ for the superficial tendons, $\mu=39.4\text{kPa}$, $\alpha=9.2$ for the deep tendons, $\mu=24.9\text{kPa}$, $\alpha=10.9$ for the flexor pollicis longus (FPL) tendon, and $\mu=12.9\text{kPa}$, $\alpha=6.5$ for the median nerve. These mean Ogden coefficients indicate that the FPL tendon was more compliant at low strains than either the deep or superficial flexor tendons, and that there was no significant difference between superficial and deep flexor tendon compressive behavior. The median nerve was significantly more compliant than any of the flexor tendons. The material properties determined in this study can be used to better understand the functional mechanics of the carpal tunnel soft tissues and possible mechanisms of median nerve compressive insult, which may lead to the onset of carpal tunnel syndrome.

Keywords

Carpal Tunnel Syndrome; Median Nerve; Flexor Tendons; Mechanical Properties

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CONFLICT OF INTEREST STATEMENT

The authors have no conflict of interest to report.

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INTRODUCTION

Carpal tunnel syndrome (CTS) is a frequently encountered chronic peripheral neuropathy caused by mechanical insult to the median nerve. The syndrome is characterized by paresthesia/tingling, numbness, and pain in the thumb, index, middle and ring fingers, and on the radial side of the palm. One plausible mechanism of CTS-provocative insult is median nerve impingement by one or more of the nine digital flexor tendons which co-occupy the tunnel. Finite element (FE) analysis provides an attractive, non-invasive means to study the potentially pathological contact stresses that may thus develop (Ko and Brown, 2007). FE simulations for such purpose ideally should be informed by physiologically-based material properties of the flexor tendons and the median nerve.

Within the extensive body of literature describing tendon mechanical properties (reviewed by Wang (2006) and by Summers and Koob (2002), among others), the vast majority of studies focus on longitudinal tensile behavior, which is the usual functional modality of tendon. By contrast, only a very few investigations have addressed transverse compressive behavior, the loading modality that would be involved in the development of contact stresses between flexor tendons and the median nerve. In work by Williams et al. (2008), a mean transverse compressive modulus of 19.49kPa was found for rabbit patellar tendons tested in unconfined compression. Using indentation testing, Shin et al. (2008) found higher compressive moduli for canine hindpaw intrasynovial tendons than for extrasynovial tendons. Lee et al. (2000) demonstrated different local compressive stiffness on the bursal versus articular sides of the supraspinatus tendon.

A similarly small group of studies has examined the transverse compressive mechanical properties of peripheral nerves. In a pair of studies utilizing *in situ* radial compression testing, the Young's modulus of diabetic nerves was found to be approximately two times greater than normal, with the diabetic nerves having a viscoelastic stress relaxation response that required a longer relaxation period to reach equilibrium (Chen et al., 2010a; Chen et al., 2010b). Similarly, parallel and radial compression tests used to estimate the transverse Young's modulus of rabbit sciatic nerves have yielded values of 41.6kPa and 66.9kPa, respectively (Ju et al., 2004; Ju et al, 2006).

These various transverse compression studies have employed differing measurement techniques for the respective specimens involved, all of whose geometries and anatomical function differ substantially from those of the digital flexor tendons and median nerve of the human carpal tunnel. The purpose of the present work was to ascertain transverse compressive material properties for human digital flexor tendons and for the median nerve. Specimen-specific FE modeling was used in conjunction with an optimization routine, allowing for inference of effective transverse material properties based on matching experimentally measured load/deformation curves.

METHODS

Experimental Testing

The nine digital flexor tendons and the median nerve were dissected from each of ten normal (thawed) fresh-frozen cadaveric forearms, ranging in age from 30 to 65 years old (5 male and 5 female). The nine tendons were the flexor pollicis longus tendon (FPL, which runs to the thumb), plus four superficial (S) and four deep (D) flexor tendons (labeled 2, 3, 4, and 5, which run to the index, middle, ring, and little fingers, respectively). Before the removal of each of these structures from the hand, the tissue sector located within the boundaries of the carpal tunnel (i.e. between the pisiform and the hook of the hamate) was marked with sutures. Accounting for longitudinal sliding motion of the tendons through the tunnel

required marking the distal sector boundary with the fingers fully flexed, and the proximal sector boundary with the fingers fully extended. For the nerve, both boundaries of the tunnel sector were marked with the fingers fully extended. A 12cm tissue length centered on the bounded tunnel sector (average 5cm in length) was used for experimental testing.

The cross-sectional area of each tendon and nerve was determined using a purpose-developed go/no-go gauge (Figure 1) comprised of a set of circular openings with diameters increasing in 0.25mm increments. Hollow cylindrical Delrin collars matching the specimen's equivalent circular diameter were then placed on each side of the testing region, adjacent to where the tissue was gripped for testing. Collars that were applied to the median nerve were split along their length to ease application without damaging the (presumed fragile) tissue. Specimens were then mounted in a specially designed transverse testing apparatus (Figure 2), using two serpentine clamps. The Delrin collars prevented tissue flattening at the clamped ends from extending into the central testing region.

As a result of the native tissue structure (distribution of collagen fibrils and matrix material) and the Delrin collars, the physical cross-section of each specimen within the apparatus was approximately elliptical. Prior to compression testing, each specimen's major axis was assessed within the testing apparatus. The tendon (or nerve) was placed under an initial axial load of 5N (or 3N, respectively) using a pneumatic cylinder (Airpel E16 D1.0, Airpot Corp., Norwalk, CT, USA), controlled by a precision regulator (MGA-100, SSI Technologies, Inc., Janesville, WI, USA). The same two convex Delrin platens (10mm in diameter) that were subsequently used to compress the tissue were driven together in 0.1mm increments by an MTS 810 actuator (under displacement control) until incipient load uptake was observed. The distance between the two convex platens was recorded as the major axis dimension of the specimen. The minor axis dimension of the tissue cross section was calculated by dividing four times the circular cross-sectional area by the quantity pi times the tissue major axis dimension.

Main compression testing was performed with the tendon (or nerve) under axial loads of 15N (or 5N, respectively). The 15N axial loading condition used for the flexor tendons was based on an isometric pinch activity with a fingertip force of roughly 5N (Dennerlein, 2004). The median nerve is likely subject to at least modest longitudinal loading associated with friction from the flexor tendons, but it is not explicitly loaded. Therefore, the nerve specimens were tested under a reduced axial load (5N).

Transverse compression of the tendon and nerve specimens was applied horizontally via movement of a platen/load cell construct mounted on ultra-low friction bearing slides (N-2 Bearing Slides, Del-Tron Precision, Inc., Bethel, CT, USA). The "floating platen" aspect of this apparatus allowed for the compressive displacement to be applied equally to each side of the tissue structure, thus eliminating any bow-stringing deformation of the tissue during testing. Transverse compression was applied quasistatically (0.5mm/sec), up to a total compression of 40% (Williams et al., 2008) of the specimen's major axis dimension. This selected compression was chosen based on approximate unidirectional transverse strains observed *in vivo* during functionally loaded hand activities, found to be in the range of 25%–40% (Appendix). Compressive load uptake was measured at a sampling rate of 1000Hz with a 111.2N (25lbf) load cell (Sensotec 3167-25, Honeywell, Columbus, OH, USA).

Compression tests were repeated three times per specimen, with a seven- or ten-minute recovery interval between tests for the tendons or the median nerve, respectively. The difference in load at a tissue's maximum compression was calculated between each of the three individual curves and the mean of those curves. The largest difference between any single curve and the respective tissue mean was 17%, although the average difference in

load at maximum compression was 3.2%. Formal comparisons of similarity in curve shape were made by calculating a sum of the squares error value relative to the average of the three curves, and normalizing that value to the maximum force and percent tissue strain. The maximum observed variation between the curves was 3.6% of the total load per 1% of tissue strain. On average, this value was much lower, typically less than 0.2% of the total load per 1% of tissue strain. Given this minimal variability between repeated tests, the data set having the smoothest force/displacement curve was selected for use in the subsequent finite element analysis.

Finite Element Modeling and Optimization

Specimen-specific FE models corresponding to each individual physical test (Figure 3) were run in ABAQUS v6.9-1 (Simulia, Providence, RI, USA). The convex cylindrical compression platens were modeled in ABAQUS CAE as rigid surfaces. Tissue specimens were modeled as prismatic, with an elliptical cross section defined using the major and minor tissue dimensions determined during experimental testing.

The ends of the tissue mesh were constrained against axial displacements during the analysis, consistent with immobilization in the serpentine clamps. Two nodes on each end of the tissue mesh were also held fixed in the transverse plane to preclude any rotation of the mesh during transverse compression. The FE transverse compression was displacement-driven, with the corresponding reaction force reported by rigid body reference nodes at the center of each platen. A first-order Ogden hyperelastic constitutive definition was used for the tendons and the nerve. For this constitutive model, strain energy W is a function of deviatoric principal stretch (λ_n):

$$W(\lambda_1, \lambda_2, \lambda_3) = \frac{\mu}{\alpha^2} (\lambda_1^\alpha + \lambda_2^\alpha + \lambda_3^\alpha - 3)$$

Here, the coefficients μ and α predominately reflect the low-strain slope and the higher-strain curvature, respectively, of the resultant stress-strain curve.

The specimen-specific FE models were used in conjunction with an automated least-squares nonlinear Matlab optimization routine (The Mathworks, Natick, MA, USA) to iteratively determine the μ and α coefficients best replicating the experimental transverse compressive behavior of each tissue specimen (Figure 4). To begin, two provisional values for the Ogden hyperelastic coefficients were loaded into an initial FE input file, run in ABAQUS, and the transverse force/displacement data from the analysis were extracted from the FE output files. FE results were interpolated so as to have force values registered at identical displacement values as recorded during the actual physical test. A least-squares nonlinear optimization function then made incremental adjustments to the Ogden coefficients to reduce the sum of the squared error between experimental and interpolated FE force values. The optimization routine equally weighted the curve fit in the higher strain and in the low strain regions. The above process was repeated until one of two optimization convergence criterion was met. Either the iterative changes in the coefficient values, or the changes in the squared error of computational versus experimental forces, decreased below the termination value of 1×10^{-4} .

Resulting Ogden coefficients were compared among the different flexor tendons and with the median nerve using linear mixed-model analysis for repeated measures, with tendons and nerve as the fixed-effect repeated measures factor. Based on estimates from this fitted model, Tukey's test was then performed to check for pairwise mean differences in Ogden coefficients between tendons and nerve.

RESULTS

Stress distributions in the tissue were non-uniform, with the maximum stress occurring at the location where the tissue was most compressed (Figure 5). The primary goal of this work was to obtain suitable material property definitions for the flexor tendons and the median nerve, and the first-order Ogden hyperelastic model gave visually acceptable fits to the experimental data using a minimal number of coefficients (Figure 6). The optimization routine yielded well-convergent Ogden coefficients. The mean Ogden coefficients (Table 1) for the superficial (S) tendons were $\mu = 35.3$ (95% CI: 26.7, 43.9) kPa and $\alpha = 8.7$ (95% CI: 6.5, 10.9). These did not differ significantly from the deep (D) tendons mean coefficients of $\mu = 39.4$ (95% CI: 29.0, 49.8) kPa ($p = 0.79$) and $\alpha = 9.2$ (95% CI: 7.2, 11.2) ($p = 0.60$). The FPL tendon had a mean μ coefficient of 24.9 (95% CI: 18.0, 31.8) kPa and α coefficient of 10.9 (95% CI: 7.7, 14.2). Compared to the superficial and deep tendons, the FPL tendon had a significantly smaller mean μ coefficient ($p = 0.040$ versus S; $p = 0.011$ versus D), but no significant difference in the mean α coefficient ($p = 0.28$ versus S; $p = 0.47$ versus D).

There were three tendon specimens (S2, D2, and D3) from the same cadaver forearm that had extremely high values of μ , perhaps indicating some undetected pathology. Excluding these “outlier” values from the calculation gave mean μ coefficients of 33.1 (95% CI: 24.7, 41.4) kPa for superficial and 35.5 (95% CI: 28.6, 42.2) kPa for deep tendons. These means were slightly smaller than the original values (which included all specimens), yet these new means had a similar pattern of difference compared to the FPL and the nerve.

The nerve was less stiff than any of the tendons. The mean nerve μ coefficient (12.9; 95% CI: 9.1, 16.8 kPa) was significantly smaller compared to the superficial ($p < 0.0001$), deep ($p < 0.0001$), or FPL ($p = 0.001$) tendons. The mean α coefficient of the nerve (6.5; 95% CI: 3.8, 9.1) was smaller than for any of the tendons, but it was only found to be significantly different from that of the FPL ($p = 0.033$ vs. FPL; $p = 0.25$ vs. S; $p = 0.27$ vs. D).

DISCUSSION

The first-order Ogden hyperelastic material property definition provided a reasonable means to describe the nonlinear transverse compressive behavior of digital flexor tendons and median nerve, up to transverse compression of nominally 40%. The tendons were, on average, approximately twice as stiff as the median nerve. In post-hoc evaluations, there was no detectable variation in tissue diameter measurement, visual appearance, or material properties associated with cadaveric specimen gender, age, weight, or height.

Somewhat surprisingly, there were only modest differences between the Ogden hyperelastic coefficients of the superficial versus deep tendons, despite striking visually-apparent differences in tendon morphology (Figure 7). Specifically, the axially oriented fibers comprising the superficial tendons appeared effectively fused into a continuum. In contrast, the deep tendons appeared as flatter, more composite structures, with individual fiber bundles evident between matrix connections. The transverse compressive behavior of these tissues is related to compression of axially-tensioned fiber bundles, which in this work were subjected to identical axial loads. A possible explanation for the similarity of tendon transverse properties, despite a dissimilar appearance, is that the transverse compressive behavior measured was more a result of bending identically tensioned fiber bundles than of compressing inter-bundle matrix material.

The FPL tendon had a lower mean μ and a higher mean α coefficient than the other flexor tendons. *In situ*, the FPL tendon bends extensively around the carpal bones towards the thumb. When excised from the hand, the natural bends and twists remain (Figure 7), indicating a fiber orientation that is not uniformly axially-directed (Wilson and Hueston,

1973). The decreased low-strain transverse compressive stiffness of the FPL tendon relative to the other flexor tendons may have resulted in part from transversely compressing FPL tendon fibers that were not equivalently/uniformly tensioned at the nominal 15N axial load, due to their non-axial orientation. The higher α coefficient may be a result of delayed load uptake by the more axially oriented fibers located deeper in the tendon (Wilson and Hueston, 1973).

There was more variation in the coefficients between individual deep tendons than between individual superficial flexor tendons, and more variation in the μ than in the α coefficient. The increased variability of the deep flexor tendon properties may have in part resulted from the difficulty in separating the small interconnections linking some individual tendons (Kilbreath and Gandevia, 1994). Sectioning of these interconnections may have increased the variability in overall cross-sectional profile, and therefore in force/displacement behavior. The larger variability in the μ than in the α coefficient resulted from the challenge of identifying the precise instant of load uptake in these very compliant tissues.

Recognizing that future investigators may have reason to model the tested structures with material property definitions other than first-order Ogden hyperelasticity, raw load/deformation data for individual specimens closest to the mean behavior for the four specific tissue types (median nerve, FPL tendon, deep tendon, superficial tendon) are provided in Supplemental Materials (electronic).

There are several limitations to this study. Rather than being perfectly prismatic, the tendon and nerve specimens had moderately non-uniform cross-sectional areas along their lengths. To minimize the effect of this variable, cross-sectional area measurements were taken as close to the compression testing region as possible. Several parametric FE investigations (Appendix) demonstrated that variability in Ogden coefficients associated with plausible uncertainty in the geometric and measurement assumptions was small (<14% for μ and <4% for α). Such a minor increase or decrease in either the μ or α coefficient would result in coefficients well within the standard deviations found for the per tissue average (Table 1). Additionally, such plausible uncertainties in the average coefficients would be insufficient to account for differences between properties of the median nerve and the digital flexor tendons (nerve μ is 64% of tendon μ , and nerve α is 29% of tendon α).

The transverse material properties determined in this study are specific to tendons under 15N of axial load. In future work, it may be informative to combine both axial and transverse behavior of the tissues, perhaps using fiber-reinforced nonlinear material models. This would involve an appreciable increase in algorithmic complexity because the optimization routine would need to accommodate the influence of both axial and transverse behavior to determine the appropriate coefficients.

In conclusion, this study presents a combined experimental/computational assessment of the apparent transverse compressive properties of the digital flexor tendons and the median nerve. As expected, the median nerve was found to be substantially less stiff than the neighboring digital flexor tendons. This information is an important foundation for further investigation of the soft tissue mechanics within the carpal tunnel, and ultimately for better understanding of the development of carpal tunnel syndrome. The material coefficients determined in this study can be implemented in anatomic finite element models of the carpal tunnel, which in turn can be used to assess the contact stresses in the median nerve resulting from impingement by the surrounding digital flexor tendons. Quantification of this plausible nerve insult mechanism could provide new insight into the pathogenesis of carpal tunnel syndrome.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

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Figure 1. Use of a purpose-developed go/no-go gauge to measure a nerve cross sectional area (a). During measurement, each tissue was confined to a circular cross section. Numeric values indicate circular diameter in millimeters, and the full range of measurements was split onto two smaller devices for easier handling (device 'a' and device 'b'). The area of the smallest circular hole that allowed unobstructed passage of the tissue was the tissue's cross-sectional area.

Figure 2.

Transverse compression testing device, with tendon specimen mounted in the serpentine clamps. During testing, the Delrin collars prevented the flattening associated with clamping from extending into the central testing region (inset). Axial load was applied by a pneumatic cylinder, controlled by precision regulator. Transverse compression was applied via convex Delrin platens, with their displacement driven by an MTS actuator. The convex platen geometry prevented stress concentrations, and facilitated alignment. All testing was performed with the tissues submerged in a saline bath maintained at 37 °C.

Figure 3.

FE model corresponding to the experimental set up, with convex circular platens, and with the tissue cross-section approximated as an ellipse. The platens were meshed with 2,500 rigid body (R3D4) elements per platen. The tendon and nerve structures were meshed with hybrid hexahedral continuum elements (C3D8H) using TrueGrid (XYZ Scientific Applications, Inc., Livermore, CA, USA), with increased mesh refinement in the contact region (inset). The number of elements in the specimen portion of the model ranged from roughly 20,000 to 30,000, depending on cross-sectional dimensions and length. The symmetry plane shows the division for the half-tendon model that was used to reduce run times in the optimization routine.

Figure 4.

Transverse force versus displacement data for multiple iterations of the optimization routine. The experimental curve is represented by the light-colored (blue) line, and the individual FE iterations are represented by the numbered dark lines. Initial values for the Ogden coefficients ($\mu = 40$ kPa and $\alpha = 12$) are used in the first FE iteration (1). Small changes are made to each individual coefficient while holding the other constant fixed (2 & 3) to evaluate the error between the experimental and FE data associated with each individual coefficient. The optimization program then used this information to make further adjustments to both coefficients (4) before arriving at the final solution of $\mu = 32.3$ kPa and $\alpha = 9.4$ (5).

Figure 5.

Compressive stress distribution for an illustrative deep (D3) digital flexor tendon (left). A compressive stress versus strain curve is shown for the selected node (*) which underwent maximal compression during the analysis (right). A half-tissue model is used to speed optimization convergence, and therefore the strain curve shown is for a node that is compressed half of the total 40% strain.

Figure 6.

Experimental transverse force versus percent transverse compression data (dark solid lines) and optimized FE results (light dashed lines) for illustrative digital flexor tendon and median nerve specimens. The flexor tendon shown is a D3 tendon with optimized coefficient values of $\mu = 32.3$ kPa and $\alpha = 9.4$. The median nerve tissue has coefficient values of $\mu = 13.1$ kPa and $\alpha = 2.0$ from the optimization routine.

Figure 7.

Illustrative appearance of superficial, deep, and FPL tendons, and the median nerve. Note the more homogenous appearance of the superficial tendon, versus the individual fibers apparent in the deep tendon. The natural curvature of the FPL is visible. There were no visible axial fibers in the median nerve.

Summary of Ogden coefficients calculated for the median nerve, flexor pollicis longus (FPL), individual superficial (S), and individual deep (D) flexor tendons (averaged from n=10 cadaver hands). The fingers are referred to numerically with index, middle, ring, and little fingers being numbered 2, 3, 4 and 5, respectively. The superficial little finger tendon (S5) was not tested due to its very small cross-section (< 2 mm diameter in most cases).

Table 1

	μ (kPa)				α			
	Average	St Dev	Max	Min	Average	St Dev	Max	Min
S2	39.7	22.6	88.4	9.9	7.7	3.6	12.6	2.6
S3	32.6	9.9	44.5	14.6	7.8	4.0	14.9	3.7
S4	33.7	13.0	49.7	5.6	10.0	4.2	17	4.3
D2	43.3	27.9	101.5	12.9	11.8	4.8	19.1	5.6
D3	47.0	28.8	125.1	24.4	7.7	4.1	15	1.8
D4	27.7	9.8	44.9	15.9	8.0	5.4	17.3	2.4
D5	39.5	11.4	56.9	19.8	9.2	4.3	17.2	1.8
FPL	24.9	9.8	39.8	11.1	10.9	4.4	16.4	3
Nerve	12.9	4.8	23.8	6.4	6.5	4.9	15.2	1.7