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Bi-directional mechanical properties of the human forearm interosseous ligament

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Abstract

Interest in reconstruction of the interosseous ligament (IOL) of the forearm in the setting of longitudinal radio-ulnar dissociation has increased in recent years with hopes of improving clinical outcomes. This increased interest has been accompanied by research on biomechanics of the IOL. However, little is known about stress and strain in the IOL under externally applied forearm loads. This information is needed to help guide reconstruction. Mechanical properties of the IOL are needed to properly model the IOL for analyses such as finite element models. The objective of this study was to document the bi-directional mechanical properties along the fiber direction (longitudinal) and perpendicular to the fiber direction (transverse). Twenty specimens were mounted in a materials testing machine to perform preconditioning and a load to failure tensile test in each direction. Strain markers on the surface of the specimens were tracked with a video system. Data analysis provided stress–strain curves for each specimen. The elastic moduli of longitudinal and transverse specimens were 515 ± 277 and 1.82 ± 2.93 MPa, respectively. The tensile strength and ultimate strain of longitudinal and transverse specimens were 54.1 ± 25.2 and 0.18 ± 0.20 and $16 \pm 5\%$ and $34 \pm 32\%$, respectively. The bi-directional mechanical properties of the IOL compared well with those published for the medial collateral ligament of the knee. The mechanical properties in the longitudinal direction were much greater than those in the transverse direction, which is indicative of the IOL's role in resisting longitudinal loading. The results of this study can be used to generate mathematical models of stress and strain in the IOL.

Keywords: Ligament; Interosseous; Properties; Tensile; Bi-directional

Introduction

The interosseous ligament (IOL) is the thickened central third ("central band") of the interosseous membrane of the forearm [5–7,13–15,17,20,23] (Fig. 1). Differential cutting studies of the IOL showed it to be the most important structural element in the interosseous membrane [5,17]. The structure is thin and broad with an average width along its insertion of 27 mm [23] and an average thickness of 1.6 mm [15]. The IOL is loaded in tension when compression is applied to the forearm, relieving load on the radial head and stabilizing the distal radio-ulnar joint [1,9,10,14,22,25,27]. With an Essex-Lopresti fracture-dislocation, IOL disruption is accompanied by triangular fibrocartilage injury and comminuted radial head fracture resulting in longitudinal radio-ulnar instability. Treatment in this setting remains challenging for the hand surgeon and has led to an increased interest in IOL reconstruction and radial head replacement [4,18,21,24].

Surgical reconstruction of the IOL is difficult because of poor access and a lack of fundamental information about the role of the IOL during forearm loading. Reestablishing performance to the level of the intact ligament remains the clinical goal, and a thorough understanding of its function and mechanical properties is necessary in developing a reconstruction. Although several studies have examined the overall load carried by

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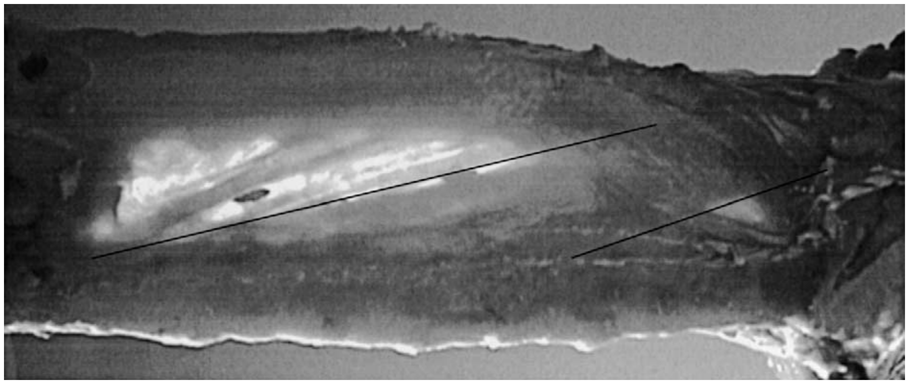

Fig. 1. The interosseous membrane with overlying tissues dissected away, held up to room light in a typical cadaver forearm. The black lines indicate the extent of the IOL.

the IOL as a function of forearm loading [1,9,10,14,22], information regarding the distribution of load in the IOL is necessary for restoration of mechanical function. Recent work demonstrated that the IOL strain distribution is non-uniform during forearm loading [8]. More information on how the IOL carries load during everyday activities and in the presence of an injury is needed to help hand surgeons make decisions regarding technique and graft choice for reconstruction.

To date, the only available information on IOL mechanical properties is for testing performed along the fiber direction of bone–ligament–bone complexes [15]. The tangent modulus of the IOL was reported as 608 ± 160 MPa, similar to the patellar tendon. Other studies of the IOL have utilized bone–ligament–bone preparations to characterize structural tensile and shear properties of the whole IOL [11,12,26].

Although these studies provide valuable information on the structural properties of the IOL, data on multiaxial mechanical properties of the IOL are necessary for the study of stress distribution in the IOL. Thus, the purpose of this study was to determine the mechanical properties of the IOL in the longitudinal (along fibers) and transverse (perpendicular to fibers) directions using specimens taken from the IOL midsubstance.

Methods

Twenty-five fresh frozen human forearms (ages 48–88) were used in this study. Each forearm was allowed to thaw overnight prior to testing. All soft tissue overlying the IOL was removed. The IOL was kept moist throughout specimen preparation with application of isotonic saline using a spray bottle. The IOL was sharply dissected from the forearm along its radial and ulnar insertions with a scalpel and placed on a light box to visualize the fiber directions. Custom made punches were used to create a longitudinal specimen with a test area of 10 mm × 2 mm (Fig. 2). Transverse specimens with dimensions of 10–15 mm × 4 mm were created using razor blades clamped in parallel (Fig. 2). Because of size constraints, a longitudinal and transverse specimen could not always be harvested from a single IOL specimen. The end result was the creation of 20 longitudinal and 20 transverse specimens from the 25 IOL specimens.

Each specimen was wrapped in isotonic saline-soaked cotton and mounted in soft tissue clamps with serrations to prevent tissue slippage. The longitudinal specimens were gripped so that the clamp-to-clamp length was 10 mm, while the transverse specimens were gripped so that the clamp-to-clamp length was 6 mm. Guides were used for the clamps to maintain appropriate alignment and gauge length. Two

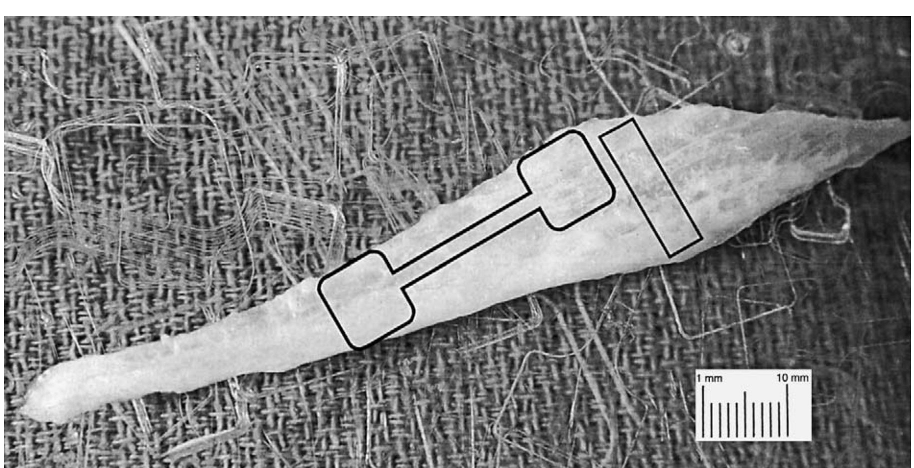

Fig. 2. Midsubstance of an IOL after excision from a cadaveric forearm with schematic showing how typical dog bone (longitudinal) and rectangular (transverse) tensile test specimens were taken.

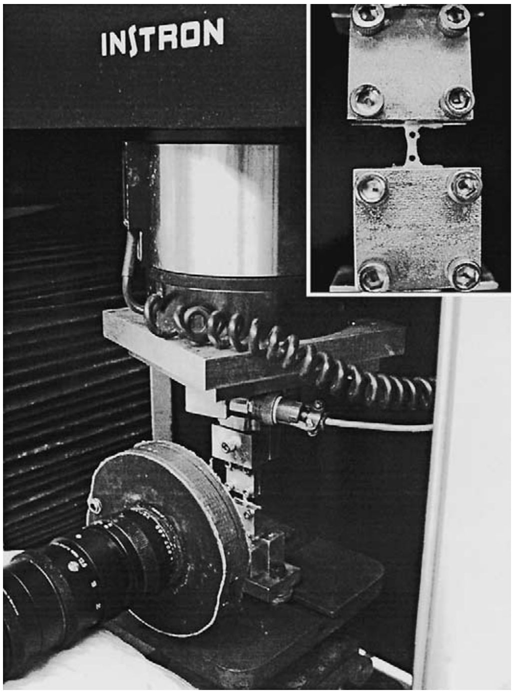

Fig. 3. Photograph of the tensile test set-up showing the video camera and a close-up view of the strain markers taken just prior to initiation of the preconditioning and load to failure protocol.

black plastic markers (1.6 mm diam, 0.3 mm thick) were affixed to the tissue using cyanoacrylate glue for subsequent strain measurements (Fig. 3). The thickness of each specimen was determined by placing the clamps on their side and taking the average of five measurements using digital calipers (Model 500-474, Mitutoyo, Kawasaki-Shi, Japan).

Specimens were mounted on a materials testing machine (Instron, Model 4502, Canton, MA). An X–Y table was used to align each specimen so that its long axis was parallel to the direction of load application (Fig. 3). Load for tests on transverse specimens was monitored using a 22 N capacity load cell (accuracy ±0.04 N, Sensotec, Columbus, OH). For longitudinal tests, a 222 N capacity load cell was used (accuracy ±0.2 N, Sensotec, Columbus, OH). Each specimen was mounted in its zero load state, preconditioned, and then loaded to failure. Because the transverse specimens were difficult to handle, they were mounted at the 6 mm gauge length, and this was assumed to be their zero load state. Transverse specimens were preconditioned with 0.1 mm elongation (1.7% clamp-to-clamp strain) at 10 mm/min for 10 cycles. Preliminary studies indicated that a preconditioning elongation greater than 0.1 mm was potentially damaging to the transverse specimens. The zero load state for longitudinal specimens was defined by repeated application of 0.5 N. Longitudinal specimens were preconditioned with 0.3 mm elongation (3% strain) at 10 mm/min for 10 cycles. Preconditioning strains were based on previous studies [28,29] and on preliminary load-to-failure tests to determine levels at which damage occurred.

After preconditioning, both transverse and longitudinal specimens were loaded to failure at 10 mm/min. Testing was performed at room temperature, and specimens were kept irrigated with application of isotonic saline using a spray bottle. During each test, load data were recorded by a computer at 10 Hz using Labview data acquisition software (National Instruments, Austin, TX). Motion of the strain markers was recorded by a video camera during the test according to established methods [28,29]. After testing, the position of the strain markers throughout each test was obtained using a Motion Analysis System (Motion Analysis™ VP320, Santa Rosa, CA). Strain for each test was determined as the change in distance divided by initial distance between the markers. The cross-sectional area of each specimen was calculated assuming a rectangular cross-section from the measured thickness and prescribed width dimensions. Load–time data, strain–time data, and cross-sectional area data were combined to obtain stress–strain curves. The tangent modulus for each specimen was calculated as the slope of the linear portion of the stress–strain curve. Tensile strength and ultimate strain were determined from inspection of the stress–strain curves. For comparison to other studies on structural properties of the IOL, the ultimate load for each test was also determined.

Statistical analysis was performed using an unpaired student's t-test to determine differences between longitudinal and transverse modulus, tensile strength, ultimate strain, and load at failure. Average stress–strain curves were generated for the longitudinal and transverse groups by calculating the mean and standard deviation of stress values at 1% strain increments up to the lowest value of failure strain.

Results

The average cross-sectional area of the specimens was 2.5 ± 0.7 mm^2 for the longitudinal group and 5.7 ± 1.0 mm^2 for the transverse group. Longitudinal specimens were much stronger than transverse specimens, failing at an ultimate load of 120.2 ± 51.4 N versus 0.82 ± 0.68 N ($p < 0.05$). All failures occurred between the strain markers with the exception of one transverse specimen. The results from this specimen were excluded from the results for average ultimate strain.

Mechanical properties of the longitudinal specimens were higher than those of the transverse specimens. The tangent modulus and tensile strength of the longitudinal specimens were over two orders of magnitude higher than that of the transverse specimens (Table 1). The ultimate strain of the transverse specimens was twice that of the longitudinal specimens ($p < 0.05$). Longitudinal specimen stress–strain curves displayed a non-linear toe region followed by a linear portion, typically followed by abrupt failure, while transverse stress–strain curves typically showed a more gradual failure (Fig. 4). Average stress–strain curves for the longitudinal and transverse directions show the large differences that were found between the two directions (Fig. 5).

Discussion

This study determined longitudinal and transverse mechanical properties of the midsubstance of the IOL. A previous study from our research center investigated longitudinal mechanical properties of the IOL using bone–ligament–bone preparations and found that the

Table 1
Average transverse and longitudinal mechanical properties from 20 tensile tests of the IOL (mean ± standard deviation) in each direction

	Longitudinal	Transverse
Tangent modulus	515.1 ± 277 MPa	1.82 ± 2.93 MPa*
Tensile strength	54.1 ± 25.2 MPa	0.18 ± 0.20 MPa*
Ultimate strain	16 ± 5%	34 ± 32%*

An asterisk denotes a statistically significant difference ($p < 0.05$) between longitudinal and transverse groups.

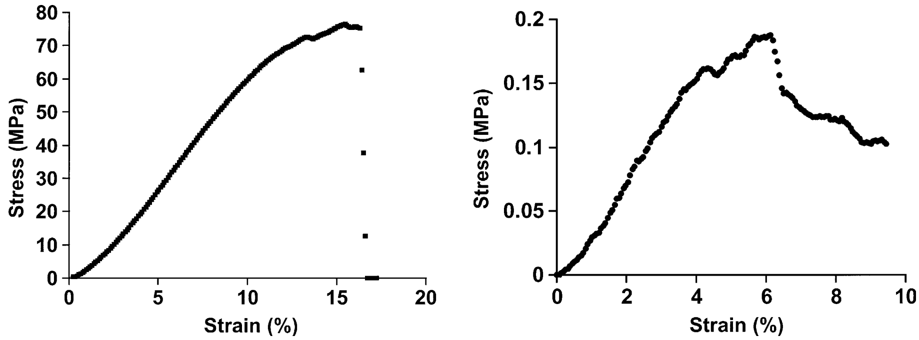

Fig. 4. Typical stress–strain curves from longitudinal (left) and transverse (right) specimens taken from the same IOL. Note the differences in stress and strain scales.

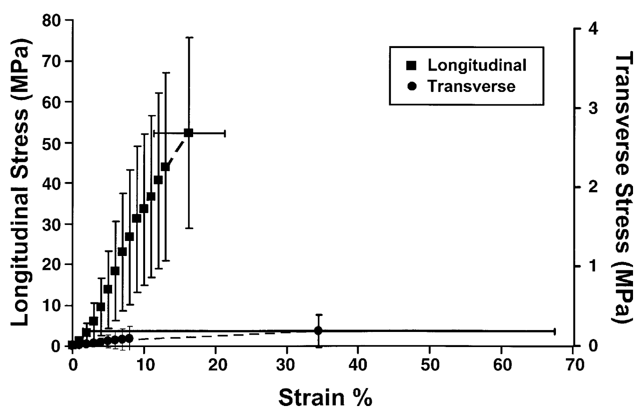

Fig. 5. Average stress–strain curves for 20 human IOL longitudinal and transverse specimens. Note the difference in stress scales (left: longitudinal; right: transverse).

average modulus, tensile strength, and ultimate strain were 608 MPa, 45.1 MPa and 9%, respectively [15]. In the current study, the modulus was slightly lower (515.1 MPa) and failures occurred at higher levels of stress and strain (54.1 MPa and 16%). These differences are likely attributable to insertion site deformation that occurs with testing of bone–ligament–bone preparation. Nonetheless, the results from the two studies for longitudinal mechanical properties are very similar.

Previous studies by McGinley, et al. determined the structural and mechanical properties of whole bone–IOL–bone preparations in response to forearm shearing (load applied along the axis of the bones) and transverse loading (load applied transverse to IOL fibers) [11,12]. Because the IOL inserts at an oblique angle to the bone, these modes of loading are complex for the IOL, with load transferred to the IOL through its insertions at an oblique angle relative to IOL fibers. The values of 3.9 and 4.7 MPa for transverse tangent modulus and ultimate strength, respectively, reported by McGinley et al. were higher [12] than those found in this study, likely due to a contribution from the IOL fiber direction. The values of 135.3 and 14 MPa reported by McGinley, et al. [11,12] for the longitudinal shear modulus and ultimate strength, respectively, were five times lower than the values found in this study for the longitudinal direction. Again, these differences may be ascribed to the mode of loading because the specimens as in this study were loaded directly inline with fibers to excised portions of IOL. Forearm shear and transverse loading through bone–IOL–bone preparations could be a relevant mode of loading for the IOL in situ, however, to determine the mechanical response at the tissue level, testing was performed along and transverse to IOL fiber direction in the current study.

Wallace, et al. reported the ultimate load and stiffness of the entire IOL when loaded in forearm shear and showed that the IOL fails at 1038 N in this mode of loading [26]. It is important to measure both structural and mechanical properties of ligaments to fully characterize their mechanical behavior. The mechanical properties measured in the current study describe the response of IOL tissue to loads applied along and transverse to the fiber direction, while the studies by McGinley, et al. and Wallace, et al. describe the response of the entire IOL to other modes of loading.

A previous study by Quapp and Weiss characterized longitudinal and transverse mechanical properties of the human knee medial collateral ligament [16]. Mechanical properties for the longitudinal direction of the MCL were slightly lower than those for the IOL in the present study. Transverse mechanical properties for the medial collateral ligament, however, were about one order of magnitude higher than for the IOL. Thus, the difference between the longitudinal and transverse modulus for the IOL was about one order of magnitude greater than for the MCL, demonstrating that transverse properties may be variable across different ligaments. Because the kinematics of the knee is different than in the forearm, the medial collateral ligament may require a higher transverse stiffness than the IOL. As the knee flexes and extends, the MCL undergoes complex loading that includes bending of the ligament about the knee flexion axis [2]. Fibers of the IOL, however, are aligned with the axis of forearm rotation [3]. As the forearm pronates and supinates, the IOL likely does not undergo bending and must bear mainly tensile loads in the fiber direction. Therefore, it is likely that the IOL has a low transverse stiffness compared to the MCL because the IOL bears mainly tensile stresses in the fiber direction while the MCL must also bear bending stresses. Other studies on tissues such as small intestinal submucosa [19] have indicated that bi-directional mechanical properties are dependent upon the degree of mechanical interactions on the fiber level. Thus, the large difference between the longitudinal and transverse fiber directions may indicate that there is less fiber interaction and cross-linking in the IOL compared to the medial collateral ligament.

Limitations of this study include variability and non-uniformity inherent in testing biologic materials. This includes the possibility of small variations in mechanical properties across the length of the specimen despite choosing specimens that were grossly uniform. Additionally, effects of metal clamps on the specimen may have caused non-uniform strain near the clamp–ligament interface. Lastly, although we were able to consistently identify the IOL, there are other fibrous bands in the interosseous membrane of the forearm which were not tested in this study and can make identification of the IOL more difficult.

The large difference in mechanical properties between the longitudinal and transverse directions demonstrates that the IOL functions mainly to carry load in the longitudinal direction and that the contribution of the transverse direction to stresses in the IOL are small. These data will help increase our understanding of the mechanical behavior of the IOL and provide data for future studies focused on modeling stresses and strains in the IOL, such as finite element models.

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