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Planar Biaxial Extension of the Lumbar Facet Capsular Ligament Reveals Significant In-Plane Shear Forces

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Abstract

The lumbar facet capsular ligament (FCL) articulates with six degrees of freedom during spinal motions of flexion/extension, lateral bending, and axial rotation. The lumbar FCL is composed of highly aligned collagen fiber bundles on the posterior surface (oriented primarily laterally between the rigid articular facets) and irregularly oriented elastin on the anterior surface. Because the FCL is a capsule, it has multiple insertion sites across the lumbar facet joint, which, along with its material structure, give rise to complicated deformations *in vivo*. We performed planar equibiaxial mechanical tests on excised healthy cadaveric lumbar FCLs ($n = 6$) to extract normal and shear reaction forces, and fit sample-specific two-fiber-family finite element models to the experimental force data. An eight-parameter anisotropic, hyperelastic model was used. Shear forces at maximum extension (mean values of 1.68N and 3.01N in the two directions) were of comparable magnitude to the normal forces perpendicular to the aligned collagen fiber bundles (4.67N) but smaller than normal forces in the fiber direction (16.11N). Inclusion of the experimental shear forces in the model optimization yielded fits with highly aligned fibers oriented at a specific angle across all samples, typically with one fiber population aligned nearly horizontally and the other at an oblique angle. Conversely, models fit to only the normal force data resulted in a broad range of fiber angles with low specificity. We found that shear forces generated through planar equibiaxial extension aided the model fit in describing the anisotropic nature of the FCL surface.

INTRODUCTION

The lumbar facet capsular ligament (FCL) is a fibrous spinal ligament that spans the articular facets of the facet joint on the posterior spine. The lumbar FCL has an innately three-dimensional shape to follow the curvature of the articular facets, and because the FCL is a capsule, it has insertion sites on more than one axis. The facet joints articulate with six degrees of freedom to facilitate and constrain the spinal motions of flexion/extension, lateral bending, and axial rotation. This combination of three-dimensional geometry, multiple attachment sites, and complex joint kinematics gives rise to complicated deformations across the FCL during physiologic loading conditions.

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Since the lumbar facet joint surface is oblique to the sagittal plane (Destouet et al., 1982; Grobler et al., 1993) spinal torsion leads to extension of one facet capsule, but lateral bending and especially flexion produce primarily shear. Thus, similar to the glenohumeral capsule (Amini et al., 2014; Gohlke et al., 1994) and the hip capsule (Pauwels, 1976), the lumbar facet capsule must be able to endure significant amounts of shear as well as extension. In light of these complex functional demands, one might reasonably expect that the fiber organization and material properties of the lumbar FCL are not uniaxially-dominated in the way that, e.g., those of the medial collateral ligament (Weiss et al., 2002; Woo et al., 1986) are.

Structurally, the lumbar FCL is marked by highly aligned type I collagen fiber bundles (Wong and Transfeldt, 2007) spanning primarily laterally between the articular facets on the posterior aspect, with some directional variation with location within the capsule (Yamashita et al., 1996). In contrast, the ventral surface (i.e., that facing into the joint space) is rich with isotropically oriented elastin (Yamashita et al., 1996). The combination of these two structurally different layers allows the capsule to endure the complex loads required for spinal function. Additionally, proprioceptive and nociceptive mechanoreceptors are found embedded in the fibrous matrixes of cervical, thoracic, and lumbar FCLs (Johnson, 2004; Kallakuri, 2012; McLain and Pickar, 1998; Suseki et al., 1997; Vandenabeele et al., 1997). These position- and pain-signaling mechanoreceptors may be activated by local deformations. Therefore, it is important to structurally characterize the lumbar FCL to aid in future studies regarding low back pain (Cohen, 2007) and degeneration (Boszczyk et al., 2003).

Mechanical testing of the lumbar FCL has generally focused on standard methods that cannot capture the full extent of the behavior, and that have been limited by the desire to produce homogeneous strain and stress fields. Little and Khalsa (2005) performed two sets of orthogonal planar uniaxial mechanical tests, in the fiber direction and perpendicular to the fiber direction. Though the lumbar FCL is never loaded under pure uniaxial conditions *in situ*, Little's data give approximate upper and lower bounds of lumbar FCL mechanical properties. Uniaxial experiments provide important insight into the anisotropy and nonlinearity of the tissue response. In this work, however, we introduce biaxial tests that provide complementary information, including coupling between the two directions and shear forces during loading. Previously, we performed equibiaxial tests on a lumbar FCL as part of a study on fitting methods (Nagel et al., 2014), but the study was primarily methodological and did not generate detailed information on the tissue's biaxial response.

Since some of the fibers of the tissue do not align with the (medial-lateral) testing axis, it is likely that the tissue generates significant shear stresses during biaxial loading (as seen in other tissue types (Fomovsky and Holmes, 2010; Sacks, 1999; Sommer et al., 2015)). To our knowledge, the in-plane shear forces and strains have never been quantified in testing of the lumbar FCL, leaving a significant gap in our knowledge of the tissue's mechanical behavior. Digital image correlation (Boyle et al., 2014; Claeson et al., 2015; Kelly et al., 2007; Raghupathy et al., 2011), however, can now provide tracking of the full displacement field, and six-degree-of-freedom (6-DOF) load cells can measure the shear forces, enabling a much fuller examination of the mechanical response of a complex tissue such as the lumbar

FCL. Thus, the objective of this work was to perform biaxial experiments on human cadaveric lumbar FCL tissue and incorporate shear forces into the analysis.

METHODS

Specimen Preparation

Six L4-L5 cadaveric lumbar facet joints (Right: $n = 2$, Left: $n = 4$) were resected from four motion segments (ages 32–65 years), cleared of all posterior musculature, and stripped of their thin surface membranes (Figure 1a). Specimens were obtained from cadavers through the University of Minnesota Anatomy Bequest Program and approved by institutional review. Spines were scanned in a 3T MRI system at the University of Minnesota Center for Magnetic Resonance Research, and samples were only taken from sites with minimal to moderate intervertebral disc degeneration (Pfirrmann grades 1–3). Facet joint capsules were visually healthy and without osteophyte or enthesophyte formation (Figure 1b). In preparation for planar biaxial mechanical testing, the trabecular bone within each articular facet of the joint was removed, and the innately curved ligament surface was flattened. The sample was trimmed to a cruciform shape, with the physiologic bone-ligament-bone attachments on one axis and ligament tabs on the perpendicular axis (Figure 1c). Ligament thickness was measured through the middle body with digital calipers.

Planar Biaxial Mechanical Tests

An Instron-Sacks planar biaxial tester, was used for mechanical characterization of the lumbar FCL samples. The machine was fitted with two dynamic load cells (500N, Instron®, Norwood, MA, USA) and two six-degree-of-freedom (6-DOF) load cells (100N, JR3 Inc., Woodland, CA, USA). The 6-DOF load cells measure one normal force, two shear forces, and three moment, though only the normal and in-plane shear forces were used in this study. Load cells were fitted such that each of the two axes had one dynamic load cell and one 6-DOF load cell. Samples ($n = 6$) were loaded with the bone-ligament-bone attachments on the horizontal axis (Axis 1) and the ligament tabs on the vertical axis (Axis 2) (Figure 1d). A preload of 0.01N was applied to each arm to measure the gauge lengths (distance from each grip to the middle of the sample) for each sample. Specimen geometries provided a longer ligament axis for all samples, with an average gauge length of $10.02 \pm 0.92\text{mm}$, compared to the average bone axis gauge length of $7.64 \pm 1.15\text{mm}$. Ten cycles of preconditioning were applied at 12% grip strain (percent gauge length) at a rate of 2.4%/3, and data were recorded at 10Hz. The four arms of the biaxial tester actuate independently, therefore, the full study protocol consisted of all combinations of one-, two-, three-, and four-arm extensions (Raghupathy et al., 2011) at 12% grip strain and 2.4%/3, so a single extension took 5s. On this time scale, we assumed that viscoelastic effects were negligible based on the observation (Little and Khalsa, 2005) that the lumbar FCL relaxed over a few minutes in uniaxial stress relaxation tests; based on Little and Khalsa's data, we would expect to see at most 5–10% relaxation in the first five seconds.

The protocol concluded with an equibiaxial extension, which was compared to the last cycle of preconditioning to ensure the sample was not damaged during testing. Data from an equibiaxial extension in the middle of the protocol were used for all further analyses;

remaining data from the extensive protocol can be used for future study. The amount of grip strain (12%) was selected to be large enough to stretch the tissue out of the toe region and produce a significant nonlinear response, but no so large as to compromise tissue integrity, based on the observations of Little and Khalsa (2005) and Nagel et al. (2014). The ligament surface was speckled with Working Elastic Stain Solution from a Verhoeff's stain kit (Sigma HT25A) for displacement tracking, and the full test protocol was video recorded at 24 frames/second.

Data Analysis

Force and Displacement—Force and displacement data from the equibiaxial extension were trimmed from approximately 50 data points to seven equally spaced points for use in the finite element model. These data points were selected to coincide with images used for displacement tracking at intervals of 2% grip strain.

Surface Displacement Tracking—Video data from the equibiaxial extension experiments were converted into sequential images. For each sample, the first (undeformed) image was imported into Abaqus CAE, and a planar mesh was created of only the ligament surface, bounded by the grips and the rigid articular facets. The mesh and the sequential images of surface deformation (spaced at 2% grip strain, consistent with the force data) were input into an image-correlation-based displacement tracking algorithm (Raghupathy et al., 2011). Surface strain maps were created using the displacement vectors at each finite element node. Displacement vectors at the grip boundaries were decomposed into their normal and shear components and averaged to verify the normal and shear displacement of the grips and bone.

Finite Element Modeling

Sample-specific FCL finite element models ($n = 6$) were constructed in FEBio (Maas et al., 2012) to fit a hyperelastic strain energy density function to experimental data. The same meshes used for surface strain tracking were converted to 3D by extruding the mesh to match the thickness ($2.65 \pm 0.30\text{mm}$) of each sample. These extruded FCL meshes contained between 1557 and 1764 brick elements. The four experimental grips were modeled as rigid bodies, and rigid interfaces were defined between the respective grips and tissue arms. Zero rotation boundary conditions were prescribed to all structures, grip displacement was constrained in the Z direction (through thickness), and the FCL mesh was unconstrained with regards to displacement. The models were displacement controlled to match experimental conditions. Normal and shear displacements were prescribed to each rigid body based on average tissue displacement at the grip boundaries found during image analysis of the experiments. The models were run through quasi-static analysis using the default BFGS nonlinear solution method in FEBio.

The ligament material was modeled in FEBio as an uncoupled (volume conserving) solid mixture containing a neo-Hookean ground matrix with the strain-energy density function

$$\tilde{\psi} = C_1(\tilde{I}_1 - 3) + \frac{1}{2}K(\ln J)^2 \quad (1)$$

where C_1 is the Neo-Hookean material coefficient, $\tilde{I}_1$ is the first invariant of the deviatoric right Cauchy-Green deformation tensor, J is the determinant of the deformation gradient tensor, and K is the bulk modulus. In addition, there were two fiber families, each described by a 2D exponential strain energy function (Equation 2) and a 2D von Mises distribution of fiber orientation (Equation 3). For the exponential fiber constitutive law (Equation 2), ξ is a measure of fiber modulus at small strain, α is the coefficient of the exponential argument, and β is the power of the exponential argument ($\xi > 0$, $\alpha > 0$, $\beta > 2$). $\tilde{I}_n$ is the square of the stretch ratio in the current fiber direction.

$$\tilde{\psi}_n(\tilde{I}_n) = \frac{\xi}{\alpha\beta} (\exp[\alpha(\tilde{I}_n - 1)^\beta] - 1) \quad (2)$$

$$R(\mathbf{n}) = \frac{\exp[b(2n_1^2 - 1)]}{2\pi I_0(b)} \quad (3)$$

The von Mises distribution (Equation 3) is an orthotropic 2D distribution in which (n_1 , n_2 , n_3) are the components of the fiber direction vector $\mathbf{n}$ (where n_1 is the preferred direction), b is a concentration parameter ($b > 0$), and I_0 is the modified Bessel function of the first kind of order 0. Additionally, a model with an isotropic elastin-rich layer was explored and found to be equivalent to the two-fiber family in a ground matrix model. Since both models were equivalent, only the fiber-matrix model was pursued.

Reaction forces at the grips were output by the model for each entered displacement, and the Matlab *fminsearch* routine was used to identify material parameters that minimized the squared error $S(\hat{\psi})$ between the model and the data. The objective function for the optimizer was

$$S(\hat{\psi}) = \sum_{f=1}^6 \left[\sum_{d=1}^6 [F_{reaction}(f, d) - F_{experiment}(f, d)]^2 \right] * W(f) \quad (4)$$

where $F_{reaction}$ are the calculated reaction forces, $F_{experiment}$ are the experimental forces, and the calculation is over six force measurements f (in-plane shear forces on each of two arms and in-plane normal forces on all four arms) at six distinct displacement points d at steps of 2% stretch along the loading curve. Out-of-plane forces were not included in the analysis. Ligament-axis normal force errors and shear force errors were multiplied by a weighting factor $W(f)$ to reduce optimization bias towards the larger bone-axis normal forces. $W(f)$ was calculated such that the peak forces across all inputs were equivalent (thus weighted as of

equal importance in the optimization) and ranged from 1 to 88. The default search tolerance of 0.01 was used, and each simulation was run with five different initial guesses for the fiber family orientations, with respect to Axis 1, to account for local minima within the objective function: $(-0.01^\circ, +0.01^\circ)$, $(-25^\circ, +25^\circ)$, $(-45^\circ, +45^\circ)$, $(-45^\circ, -0.01^\circ)$, and $(+45^\circ, +0.01^\circ)$. A sensitivity analysis of the other six parameters yielded no effect on the calculated optimal solution. To determine the effect of shear forces on the fits, the optimization was also run using only the normal forces.

Fibers were assumed to have identical material properties as defined by the strain energy parameters (α, β, ξ) but were treated as independent in terms of the preferred fiber angle (θ_1, θ_2) and von Mises concentration parameter (β_1, β_2) . A small value of b indicates a more anisotropic distribution, whereas a larger value of b describes a distribution that is aligned around the angle θ . Identical fiber material property parameters were chosen to simulate the uniform properties of collagen fibers, as we know of no evidence that suggests properties differ across fiber types. Thus, including individual material properties for each fiber family would overparameterize the problem at hand. The bulk modulus (K) in the Neo-Hookean model (Equation 1) was set to one thousand times the shear modulus (C_1) to create a nearly incompressible model. Eight parameters in total, $(C_1, \alpha, \beta, \xi, \theta_1, \theta_2, b_1, b_2)$, were thus optimized. All simulations were carried out on one core at the University of Minnesota Supercomputing Institute. For consistency, the parameter values θ_1, θ_2 of right FCLs were computationally flipped to match the orientation of left FCLs.

Following the optimization, 95% confidence intervals were computed for each parameter by the method of Draper and Smith (Draper and Smith, 1981), which yields the following expression for the confidence intervals.

$$(\psi - \hat{\psi})^T \hat{Z}^T \hat{Z} (\psi - \hat{\psi}) \leq p \left(\frac{S(\hat{\psi})}{[n-p]} \right) F(p, n-p, 0.95) \quad (5)$$

$$\widehat{Z}_{ij} \equiv \frac{\partial \varepsilon_j}{\partial \psi_i} \quad (6)$$

ψ_i is the vector of parameters, and all terms with a hat denote those related to the fitted parameter set. p is the number of parameters ($p = 8$) and n is the number of observations. For shear fits, $n = 36$ (6 forces x 6 displacement points), and for fits without shear forces, $n = 24$ (4 forces x 6 displacement points). $\widehat{Z}_{ij}$, the sensitivity of the error to each parameter, is obtained numerically by making small perturbations around $\hat{\psi}$. Although the optimization problem was non-linear, confidence intervals for each parameter were computed based on linear theory following Draper and Smith (Draper and Smith, 1981) by varying one parameter at a time and leaving the remaining seven parameters constant. The linearized form of Equation 5 is

$$S(\psi) \leq S(\hat{\psi}) \left\{ 1 + \frac{8}{[n-8]} F(8, n-8, 0.95) \right\} \quad (7)$$

To grow the confidence region bidirectionally, the parameter of interest was sequentially incremented until $S(\psi)$ was greater than the right side of Equation 7. If $S(\psi)$ was less than the right side of Equation 7 after a sixteen-fold change in parameter value, the confidence interval was specified as infinite. All computations were completed on a single core at the University of Minnesota Supercomputing Institute.

RESULTS

Mechanical Testing

Normal-force-displacement curves (Figure 2a) for lumbar FCLs in equibiaxial tension highlight the anisotropy of the tissue. Along the bone-axis (Axis 1), the FCL is much stiffer and exhibits a smaller toe region than along the ligament-axis (Axis 2). Peak normal forces averaged $16.11 \pm 4.41\text{N}$ along the bone-axis (N1), in contrast to those ($4.67 \pm 3.05\text{N}$) along the ligament-axis (N2). Out-of-plane shear force-displacement curves are included in supplementary data, though the three moments (M_X , M_Y , M_Z) are not reported, as they were negligible. The RMS error between grip displacement and average tissue displacement at the grip boundary for all grips and all samples was 0.16mm (0.19%), confirming that grip slipping was at a minimum. Furthermore, sample integrity was not compromised during the extensive protocol because last equibiaxial extension was, on average, only 4% lower than the last cycle of preconditioning.

Because of variance in tissue structure, the signs of the shear forces (Figure 2b) were not the same for all samples, and no trends existed between the shear force sign and the grip location at which it was recorded. Shear forces on the bone-axis (S1) obtained a higher peak force magnitude ($3.01 \pm 2.61\text{N}$) than ligament-axis (S2) shear forces ($1.68 \pm 0.65\text{N}$), although the two were not significantly different ($p = 0.25$). Pooled shear force magnitudes were significantly smaller than both the fiber-direction normal forces ($p < 0.0001$) and the cross-fiber-direction normal forces ($p < 0.05$).

Data Fitting – Normal and Shear Forces

The eight model parameters were regressed to the normal force and shear force data for six L4-L5 FCL planar equibiaxial extension experiments. Different initial guesses gave different fits because of the presence of local minima in the objective function. Initial guesses of $(-25^\circ, +25^\circ)$ and $(-45^\circ, +45^\circ)$ most often resulted in fits with the lowest sum of squared error. In general, error contributions from shear data were larger than error contributions from normal force data. Figures 3a–b show the experimental (dots) and model (line) forces versus stretch for the best fit of a representative sample. The model captured the different shear force signs in the two directions but did not capture the non-linearity of the shear force, S2 (i.e., the medially acting force on the inferior arm of the cruciform).

The optimal parameter values with standard deviations are displayed in Figure 3c. The average matrix stiffness, given by the Neo-Hookean term C_1 , was 0.018 ± 0.011 MPa. The prefactor, ψ , in the exponential fiber strain energy density function had the largest value (3.64 MPa) and the widest spread (2.25 MPa) of the three parameters (α , β , ψ). The exponential term β had the smallest spread with respect to the parameter value (2.57 ± 0.29), and α ; averaged 0.73 ± 0.29 . The primary fiber direction parameter (θ_1) for all samples was oriented in the superior-lateral to inferior-medial direction, defined by a negative angle, with an average value of $-22.40 \pm 8.62^\circ$. The secondary fiber direction parameter (θ_2) resulted in both positively and negatively oriented fibers. The primary fiber direction was highly aligned ($b_1 > 4.4$ for all cases). The secondary fiber direction was less strongly aligned, with a b_2 range of 0.86 to 3.86 across all samples.

Parameter values were divided by their respective 95% confidence region width to normalize each goodness of fit. Normalized values greater than one indicated a confidence region size smaller than the parameter value, and thus a good estimate of the parameter value. The exponential term β has the tightest confidence region consistently producing values greater than one. Three parameters had normalized values less than one, C_1 , ψ , and b_1 , meaning these parameters were the least specific. For b_1 , the upper 95% confidence bound was calculated as infinite for three of the six samples, generating a very large confidence region width. However, when $b > 5$, the 2D von Mises distribution is close to fully aligned. Therefore, the large confidence region for b_1 merely suggests that once the fibers are strongly aligned, there is very little effect if they become even more strongly aligned. The 95% confidence intervals for each parameter are computed in Table 1 located in the supplemental material and the normalized values are plotted with standard deviations in Figure 4.

The experimental surface strains were compared to those predicted by the model (Figure 5) by calculating the RMS error between the nodal strains. Nodal strains from all samples were pooled to calculate the RMS error in the E_{XX} (Axis 1), E_{YY} (Axis 2) and E_{XY} (shear) directions, and resulted in visually similar strain fields but errors of 7.46%, 5.74%, and 4.37% strain, respectively. These fairly large strain errors show that even this eight-parameter model may not be sufficient to capture the full behavior of the material. Possible extensions to the model will be described in the Discussion.

Data Fitting – Change in Model Fit Without Shear Forces

When the model was re-fit without using the shear data, the resulting minimum sum of squared errors for the normal forces were smaller and less varied (range $\mathcal{S}(\hat{\psi}) = [0.25 \ 2.94]$) than those found previously (range $\mathcal{S}(\hat{\psi}) = [0.52 \ 12.21]$). This result is not surprising, since the need to fit the shear force data pulled the model away from the normal force data. The primary and secondary fiber angles for both fits are shown in Figure 6, column a. Despite having the opposite sign, both fiber directions contributed to low error in the normal force fits (Figure 6, column a), but the parameters calculated using only normal force data were not capable of capturing the magnitude and direction dependence of the shear forces (Figure 6, column c). Mean parameter values for the ground matrix and strain energy density function did not change greatly when the shear forces were removed, but the fitted values

had greater variability, as indicated by larger standard deviations. Both the primary and secondary fiber directions were estimated poorly using only normal force data; the inclusion of the shear forces led to more consistent and higher-confidence direction. Figure 4b shows the effect of fiber angle shear force magnitude and direction. While b_1 and b_2 have tight confidence regions creating normalized values greater than one, the primary (θ_1) and secondary (θ_2) fiber angles, whose distributions they specify, have wide confidence regions creating normalized values less than one. RMS errors in nodal strains between the normal force only model and the experimental strain maps (E_{XX} , E_{YY} , E_{XY}) were not significantly different from the error calculated from the model inclusive of the shear forces.

DISCUSSION

We used planar biaxial mechanical testing and finite element analysis to determine the mechanical properties of the L4–L5 facet capsular ligament (FCL) in equibiaxial extension. Two actuators of the mechanical test setup were fitted with load cells capable of measuring both normal and shear reaction forces. Experimental data produced measurable shear forces during equibiaxial extension that were not significantly different in magnitude from normal forces produced along the ligament-axis (Axis 2). Model parameters based on the shear force data as well as normal force data showed less variation and identified two fiber populations, one running primarily in the superior-lateral to inferior-medial direction, and the other running roughly lateral-to-medial but with variation among samples.

Data from the current study can be compared to those from the two other studies that performed planar mechanical tests on cadaveric lumbar FCL samples. Biaxial testing of a cadaveric lumbar FCL by Nagel et al. (2014), yielded peak bone-axis normal forces ($\sim 20\text{N}$) within range of the current study. Little and Khalsa (2005) reported elastic moduli at 10% and 20% strain for orthogonal uniaxial data sets. We simulated Little's experiment for each of our material fits ($n = 6$) by applying two uniaxial deformations along the X and Y axis of a $1 \times 1 \times 1\text{mm}$ cube to 10% and 20% strain. The simulated elastic moduli along the fiber direction at 10% and 20% strain, calculated based on the results obtained in the current study, were greater ($4.34 \pm 1.25\text{MPa}$ and $7.82 \pm 3.13\text{MPa}$, respectively) than those computed by Little ($0.10 \pm 0.09\text{MPa}$ and $0.18 \pm 0.13\text{MPa}$). However, the simulated elastic moduli perpendicular to the fiber direction at 10% ($0.52 \pm 0.34\text{MPa}$) and 20% ($1.17 \pm 0.79\text{MPa}$) strain were similar to Little's values ($0.23 \pm 0.20\text{MPa}$ and $1.04 \pm 0.86\text{MPa}$). The discrepancy between the data sets may arise from the inherent differences between uniaxial and biaxial tests. In uniaxial stretch, fibers can undergo large non-affine rotations, allowing the tissue to reach much larger strains before collagen fibers become engaged (Chandran and Barocas, 2006). However, the cross coupling of fibers during biaxial extension strongly limits fiber rotation and results in more fiber stress at lower strain. The model here assumes affine deformations and may be overestimating stresses as a result.

The current study is not without limitations. Placing load cells measuring both normal and shear forces on all four actuators, instead of two, would provide a more detailed analysis of shear reaction forces. The additional shear measurements would also increase the specificity of the finite element model. The FCLs were prepared in a planar orientation for biaxial testing, however, a slight curvature, although greatly reduced from the *in situ* curvature, still

existed, and the experiment was replicated with a finite element model that assumed a planar surface. Therefore, in-plane surface strains output by the model would underestimate the experimental strains, which may account for some of the error between the experimental and model strains (3-D motion of the lumbar FCL is discussed further in Claeson et al. (2015)).

The eight-parameter hyperelastic model, that was selected to describe the lumbar FCL, assumes homogeneity across facets including fiber orientation and transition zone (ligament to bone) properties, which could also account for error between the optimization and the experimental data. Recently, our group (Zarei et al., submitted) has shown, through use of polarization sensitive optical coherence tomography and multi-scale finite element modeling, that the fiber bundles of the lumbar FCL are highly aligned but are aligned about different angles at various regions across the ligament surface. The current model, while fitting two separate fiber distributions, assumes a uniform distribution and direction, which does not account for this regional inhomogeneity. The region in which the ligament inserts into the rigid articular facets, or enthesis, was not modeled explicitly. It has been shown (Thomopoulos et al., 2006) that this region has heterogeneous properties in regards to fiber stiffness and fiber bundle orientation. Considering the heterogeneous properties of this material transition zone may be of particular importance as the model reaction forces obtained from this region drive the goodness of the overall model fit.

The current approach could be expanded to account for surface displacements and viscoelastic properties. Parameters could be tuned to the experimental surface strains, in addition to the grip forces, although the surface strains would likely not be indicative of the through-thickness deformation because of the transition from anisotropic collagen fiber matrix to an isotropic elastin-rich matrix. Additionally, the model assumed a homogenous material type and did not account for the through-thickness material transformation or for any in-plane variation in structure or properties. The importance of viscoelasticity in various spinal ligaments has been reported by others (Bonifasi-Lista et al., 2005; Little and Khalsa, 2005; Mattucci et al., 2013; Troyer and Puttlitz, 2011; Troyer and Puttlitz, 2012), and the viscoelastic behavior of the lumbar FCL would be an excellent subject for further study. The quantities reported here should be treated as instantaneous parameters because we performed relatively rapid experiments and did not account for relaxation. Many researchers (e.g., Quinn and Winkelstein, 2010; Raghupathy et al., 2011; Voycheck et al., 2014) have used polarized light imaging (PLI) (Tower et al., 2002) to quantify collagen fiber orientation for validation of finite element models. The lumbar FCL, however, is opaque and too thick (approximately 2.5mm) to allow imaging via this transmission-based method. Other modalities, such as polarization-sensitive optical coherence tomography (Wang et al., 2011, Claeson et al., 2015) or scanning electron microscopy (Iatridis and ap Gwynn, 2004; Provenzano and Vanderby, 2006) could be considered in the future. Finally, it is also emphasized that we focused on healthy tissue in the current study, but low back pain can occur with lumbar FCL degradation (Cohen, 2007; Lewinnek and Warfield, 1986), therefore, a comparison between healthy and degenerated samples would be in order. All of these possible extensions are worthy of consideration, whereas the current study focused on establishing a baseline (healthy samples, minimal viscoelastic effect, assumed homogeneity of the tissue) upon which subsequent work could build.

CONCLUSIONS

This study highlights the importance of shear forces in lumbar FCL deformation. Shear forces were measured on the same order of magnitude as the normal reaction forces on the ligament-axis, and model parameters showed more consistency when the shear forces were used in the fitting process.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

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HIGHLIGHTS

- The lumbar facet capsular ligament (FCL) has complex geometry and structure, giving rise to complex *in vivo* deformations
- Normal and shear reaction forces were recorded during planar equibiaxial extension of lumbar FCLs
- An eight-parameter anisotropic, hyperelastic model was fit to experimental data
- Including experimental shear forces in the fitting process yielded more robust parameter estimates
- The best-fit model included highly aligned fibers, indicative of the anisotropic nature of the lumbar FCL

FIGURE 1.

Lumbar facet capsular ligament (FCL) preparation for planar biaxial mechanical testing. **(a)** Schematic of the posterior lumbar spinal column from the third lumbar vertebra (L3) to the fifth lumbar vertebra (L5). The FCL (not shown) spans between the rigid superior and inferior articular facets (S/IAF) that bilaterally flank the spinous processes (SP) **(b)** Cadaveric L4–L5 motion segment in same view as (a) with posterior musculature removed to expose the bilateral facet joint capsules. **(c)** The capsule was resected from the motion segment, and the thin surface membrane was removed. **(d)** Some trabecular bone was removed to create a planar orientation. The collagen fiber bundles were preferentially oriented between the remaining rigid articular facets (BONE), and free ligament ends (LIG) were along the perpendicular axis. **(e)** The sample was loaded into grips connected to load cells in the planar biaxial setup. The fibers were aligned on the horizontal axis (Axis 1) and the ligament tabs on the vertical axis (Axis 2). Normal forces were measured from all four load cells, and shear forces were measured through load cells attached to the grips labeled INFERIOR and IAF (L4). TP: transverse process; SP: spinous process; IAF: inferior articular facet; SAF: superior articular facet

FIGURE 2.

Planar biaxial mechanical tests give (a) normal force- and (b) shear force-stretch curves for $n = 6$ samples. (a) Along the fiber axis (N1, dashed), forces reached a larger peak magnitude than did those along the ligament axis (N2, solid). (b) Peak shear forces (pooled S1 and S2) were lower than both peak N1 forces and N2 forces. Peak shear forces of S2 (solid) were of lower magnitude than those of S1 (dashed).

FIGURE 3.

Model fit (lines) to a representative experimental (a) normal force- and (b) shear force-displacement data. The model accurately captures the sign of the normal and shear forces, but cannot capture the full non-linearity of S2. (c) Optimal parameter values for the model fits for all optimization simulations ($n = 6$) with standard deviations. The parameters α and β had the smallest distribution of values. The primary fiber angle (θ_1) was distinctly negative in value and was highly aligned (b_1), whereas the secondary fiber angle (θ_2) had a larger distribution and less strength of alignment (b_2).

FIGURE 4.

Goodness of fit for each parameter obtained by dividing parameter values by their respective 95% confidence region width is shown for the **(a)** parameters fit with shear and normal force data and **(b)** parameters fit with only normal force data. Good parameter estimates have a value greater than one. **(a)** The shear-and-normal-force model fit gave the fiber angles (θ_1 , θ_2) with high confidence. The fiber distribution parameter (b_1) falls below zero because the upper 95% confidence bound was calculated as infinite for half of the samples, generating a large confidence region width. The low confidence for b_1 suggests that once the fibers are strongly aligned, there is little effect if they become even more strongly aligned. **(b)** The normal-force-only model fit the fiber angles (θ_1 , θ_2) with a low-level of confidence and instead, specified the matrix parameter (C_1) with a high confidence.

FIGURE 5.

Surface displacement maps showing displacement vectors and shear (E_{XY}) strains from a representative experiment along with the predictions from the model fit to the force data. Experiment and model fit are both at 12% equibiaxial extension. The RMS error for this representative sample is 5.29% shear strain.

FIGURE 6.

Comparison of the shear force model fit (**Row 1**) and normal force model fit (**Row 2**) for a representative sample. (**Col a**) The primary fiber angles (bold) from the optimizations fit to the opposite sign, and both secondary fiber angle fits are relatively horizontally aligned. (**Col b**) Despite the opposite primary fiber angle sign, both models capture the normal force fits as expected, but (**Col c**) the normal force model does not match the sign of the shear forces.