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Marker-Free Tracking of Facet Capsule Motion using Polarization-Sensitive Optical Coherence Tomography

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Abstract

We proposed and tested a method by which surface strains of biological tissues can be captured without the use of fiducial markers by instead, utilizing the inherent structure of the tissue. We used polarization-sensitive optical coherence tomography (PS OCT) to obtain volumetric data through the thickness and across a partial surface of the lumbar facet capsular ligament (FCL) during three cases of static bending. Reflectivity and phase retardance were calculated from two polarization channels, and a power spectrum analysis was performed on each a-line to extract the dominant banding frequency (a measure of degree of fiber alignment) through the maximum value of the power spectrum (maximum power). Maximum powers of all a-lines for each case were used to create 2D visualizations, which were subsequently tracked via digital image correlation. In-plane strains were calculated from measured 2D deformations and converted to 3D surface strains by including out-of-plane motion obtained from the PS OCT image. In-plane strains correlated with 3D strains ($R^2 = 0.95$). Using PS OCT for marker-free motion tracking of biological tissues is a promising new technique because it relies on the structural characteristics of the tissue to monitor displacement instead of external fiducial markers.

KEY TERMS

biomechanics; image correlation; polarized light; spine

INTRODUCTION

The lumbar facet joint, which allows for posterior spinal motion, consists of four structures: two rigid articular facets from adjacent vertebrae, the ligamentum flavum, and the facet capsular ligament (FCL). The FCL is responsible for guiding the relative motion between articular facets and containing the synovial fluid within the joint. During spinal flexion, the articular facets rotate in the sagittal plane, causing shear across the lumbar FCL²². The collagenous fiber bundles of the lumbar FCL are highly aligned and oriented medially-laterally between the articular facets³⁷. Thus, flexion creates a complicated strain profile across the ligament surface. In addition to complex kinematics, the lumbar FCL is thick,

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approximately 3 mm, and has an innate convex curvature. Capturing the complex, three-dimensional surface kinematics of the FCL *in situ* during spine motion is extremely difficult.

In light of this challenge, two primary approaches have been taken to track FCL kinematics: bone-based and marker-based. The bone-based approach involves imaging vertebral motion and then inferring FCL kinematics. The availability of single-fluoroscopic²⁴ or dual-fluoroscopic¹⁸ measurements makes this approach attractive, in particular when combined with other imaging modalities and finite-element models, but it is still an indirect measure of FCL motion. The marker-based approach¹⁶ involves suturing markers into the capsule to allow tracking of material points. The marker-based approach gives a direct measure of the FCL motion but is limited to a finite number of points and thus a relatively coarse measurement.

Optical coherence tomography (OCT) is a non-destructive, depth-resolved, back-scattering/reflection-based imaging technique that has been used previously in mechanical investigations of wound healing⁹, tissue folding¹⁴, and thermal denaturation of collagen^{2, 11}. Importantly, Filas et al.^{13, 14} used OCT images of the embryonic chick heart with microspheres as markers to track tissue deformation, and Li et al.²⁰ used OCT for gross strain and strain rate estimates based on heart wall thickness changes. Although neither of these approaches would be directly applicable to the problem at hand, they demonstrate the potential of OCT as a deformation/strain imaging tool. A reflection-mode technique is essential for optical imaging of the lumbar FCL because the FCL is a thick and highly scattering sample for imaging by conventional transmission-mode techniques such as polarized light imaging (PLI)^{34, 35}, which has been used on the thinner cervical FCL^{19, 26, 27}.

Polarization-sensitive (PS) OCT is advantageous over standard OCT in that it measures the birefringent properties of biological tissues as an additional contrast. PS OCT captures accumulation of retardance over depth, and optic axis orientation for fiber alignment, as characteristics of anisotropic tissues. Retardance and axis orientation measurements from PS OCT have been used previously to identify fiber pathways in the brain³⁶ and fiber alignment in tissue-engineered tendon¹.

It may be possible to track 3D structural reorganization with PS OCT because the system can record changes in surface geometry and can identify innate structural characteristics with micrometer-scale resolution. These structural features could serve as fiducial points to track motion, as done previously with transmission-based polarized light imaging^{8, 28}. If tracking can be combined with the surface geometry acquired routinely in OCT imaging, then the full 3D displacement field of the tissue surface can be obtained, allowing the methods of Filas et al.¹³ to be used to calculate surface strains. This study aims to utilize PS OCT, without the aid of fiducial markers, to extract three-dimensional surface strains in the FCL during stepwise bending of an isolated motion segment.

METHODS

Specimen Preparation

Four cadaveric L3–L5 motion segments, ages 29 – 61 years, were obtained from the University of Minnesota Anatomy Bequest Program for use in this study. Of the eight available L4–L5 FCLs on the four motion segments, three were excluded because of gross collagen fiber damage and/or osteophyte formation near the joint space. Three left FCLs and two right FCLs ($n = 5$) were included in the study and showed no visible signs of structural or kinematic degeneration in spite of the range of specimen ages. One bilateral pair of FCLs was from the same motion segment, and one FCL was imaged twice to check reproducibility.

Motion segments were cleared of all posterior musculature, and intervertebral discs were removed at the pedicles, leaving only the posterior elements intact. The spinous processes were also resected to create more space for focusing the OCT beam. To impose reproducible vertebral body motion, three bone screws were drilled into the L3, L4 and L5 laminae of each specimen. The preparation is shown schematically in Figure 1. The rigid bone screws, when squeezed with towel forceps, induced sagittal plane rotation analogous to motion segment flexion. One ‘click’ during forceps depression displaced the clamps 3 mm, corresponding to roughly 1.87° of relative rotation between the vertebrae. We note that the motion was not intended to reproduce the actual motion during flexion, which depends on the action of the spinal muscles and connective tissues, but rather to provide well-controlled, reproducible bending of the motion segment. For comparison, Kozanek et al.¹⁸ report *in-vivo* sagittal plane L4–L5 facet joint displacements of approximately 2 mm from full flexion to full extension in healthy subjects. We expected a lesser amount of FCL displacement in the current study because the relative rotations we induced were only flexural. Additionally, the motion segment’s rotation axis was moved posteriorly from the removal of the intervertebral discs, which created less relative motion through the facet joint.

The L5 vertebra of each specimen was immobilized via a vice grip. Video was captured while the towel forceps clamped the L3 and L5 bone screws. Figure 2 shows the L5 facet joints of a L3–L5 motion segment fixed within the vice grip, and the method by which relative rotation was achieved. The video was imported into ImageJ³⁰, and the bone screw heads and insertion points were tracked over the duration of the image sequences. Imposed motion was quantified by subtracting the initial angle between the bone screws from the angle between the clamped bone screws during the sequence. The relationship between a forceps ‘click’ and the induced bend angle were recorded for each specimen.

Experimental Protocol

For imaging the samples, we used a polarization-maintaining-fiber-based swept-source polarization-sensitive optical coherence tomography (SS-PSOCT) system. The center wavelength was at 1300 nm with a full width at half maximum bandwidth of 78 nm. A detailed description and characterization of the optical setup has been previously reported with several modifications in this study³. Briefly, the interference signals on the main and crossed polarization channels are split by a fiber polarization splitter and detected by

balanced detectors. The a-line scan rate of the system was 42 kHz and the signal was recorded at a sampling rate of 250 MS/s and an anti-aliasing filter with a cut-off frequency of 50 MHz was applied. Data were acquired with a high-performance digitizer (NI-5761, National Instruments, Austin, TX) and the implementation of the anti-aliasing filter and data pipelining, making high-throughput real-time streaming possible, were accomplished with custom programming of an FPGA module (NI 7965R). Both the digitizer and FPGA module operated on the same PXI platform. This resulted in an imaging range of 2.6 mm , which can be increased up to five times using the full throughput of 250 MS/s . This imaging range was sufficient for the FCL sample since penetration was no greater than one mm . The beam spot on the sample had a $1/e^2$ diameter of 48 μm resulting in a Rayleigh range of 1.4 mm . The galvanometer scanner scanned the sample laterally in two dimensions ($x - y$, raster scan), which allowed the *en-face* and contour analysis, whereas the information along the depth direction (z) is provided by the inverse Fourier Transform of the raw spectral data. Vibrations from the raster scan were negligible because the high-precision galvanometer used raised the 1 nm noise floor of the system by only 0.5 – 1 nm , thus altering the retardance degree by 0.0000055°.

The specimen was placed on a custom stage under the scanning lens of the SS-PSOCT with the posterior side facing up (Figure 3A). The stage was bolted securely to the tabletop to prevent extraneous motion during imaging. A thin slit was cut into the surface of the stage to allow the bone screws to protrude downward into the stage cavity while the motion segment lay flat on the stage (Figure 3B). The specimen was secured to the stage, ensuring that in-plane motion over the FCL surface was only a result of the imposed bending and not from gross translation of the motion segment. A window was cut into the side of the stage, permitting the towel forceps access to the bone screws in the stage cavity. Polymer retarder film (Edmund Optics, Barrington, NJ) was mounted onto a slide and attached to a lateral edge of the imaging window. The edge of the film was trimmed in an irregular, jagged orientation to facilitate image registration between resulting image stacks.

A $9.0 \times 7.2 \times 1.37$ mm volumetric imaging window was captured including 250 c-lines (frames) at increments of 29 μm , which allowed for depth information 1.1 mm through the tissue thickness. Imaging began at the lateral aspect of the FCL, traversed the FCL surface, and ended at the medial aspect (Figure 3C). Since the right FCL is the mirror image of the left, the data were computationally manipulated so that in all cases the right side of the frame was the superior FCL and the left side of the frame was the inferior FCL (the natural anatomical orientation of a left FCL).

At the time of testing, the motion segment was positioned with the L4–L5 FCL of interest under the SS-PSOCT beam. The motion segment was then secured to the testing stage and hydrated with 0.9% saline solution. The towel forceps were clamped around the screws but not squeezed. The beam was focused by adjusting its vertical position with respect to the FCL surface. The film was positioned at the edge of the imaging window such that approximately 100 a-lines were included (10% of a cross-sectional frame). Once the imaging area had been confirmed and the initial image (REST) had been focused, the volumetric scan commenced, and the data were captured within a few seconds. For the first bend case (BEND 1), the towel forceps were then depressed to a screw angle change $\sim 5^\circ$

(approximately half the range of maximum L4–L5 motion segment flexion²⁴), the initial image was refocused, and the volumetric scan was captured. This sequence was repeated for BEND 2 at $\sim 10^\circ$ (maximum L4–L5 motion segment flexion). The sample was only hydrated once immediately before capturing the initial scan because each scan was completed within seconds. Water does increase scattering and can reduce the image intensity, but if the signal is strong, the effect is minimal.

Data Processing and Image Analysis

To construct the images from the data, the spectra on two orthogonal polarization channels were interpolated and resampled from the wavelength domain to k-space using the grating equation¹². After residual background from the reference arm had been subtracted, dispersion mismatch between the sample and the reference arm was compensated for numerically⁷. The side lobes were suppressed by apodization using a Hamming window. Lastly, the inverse Fourier transform was applied to the spectra in k-space that carried the complex signals in depth.

For the two polarization channels, the amplitude A_1 and A_2 were extracted. The reflectivity (R) and phase retardance (δ) were calculated as,

$$R(z) = A_1(z)^2 + A_2(z)^2 \quad (1)$$

$$\delta(z) = \tan^{-1}(A_1(z)/A_2(z)) \quad (2)$$

where z denotes the depth. The low signal-to-noise regions of the reflectivity and phase retardance images were masked for further data processing. A mask was created using segmentation to remove noise from each cross-sectional reflectivity image, and a median filter further reduced the speckle noise. The mask was applied to the initial reflectivity and retardance images.

Three sets of image stacks (REST, BEND 1 and BEND 2) were acquired for each L4–L5 FCL. Overlapping volumetric areas were identified and registered; extraneous frames and a-lines were removed from analysis, thus the full $9.0 \times 7.2 \text{ mm}$ imaging area was not analyzed. The images were registered by watching the irregular edges of the film enter and exit the imaging window. Between 40 and 75% of the volumetric area remained for analysis after image registration (mean $60.2 \pm 13.0\%$).

Figure 4 displays the image processing scheme for a representative a-line (phase retardance) within a frame in an image stack. The banding pattern is due to phase wrapping. Each image stack (Figure 4A) was filtered in the two transverse directions with an ideal low-pass filter at 20 pixels^{-1} , and the signal length within each a-line (unfiltered dimension) (Figure 4B) was computed. Signal length was used to translate the data into physical units; an a-line composed of 219 pixels is equivalent to 1 mm of depth penetration through the lumbar FCL. Unfiltered a-lines from the filtered image stack were then transformed into Fourier space, and a spectral analysis was performed on each a-line (Figure 4C). The maximum value of the power spectrum (related to the uniformity of birefringence along the depth direction) at

each a-line was recorded and its location was used to determine the dominant banding frequency (a measure of degree of fiber alignment) for all a-lines in the image stack. The dominant banding frequencies and the maximum values of the power spectra were mapped to separate 2D arrays (Figure 4, Row D).

Motion Analysis

Two types of *en-face* images of the lumbar FCL surface were reconstructed to determine the best set of tissue characteristics to use to track motion of the FCL between cases of static bending. Figure 5 shows reconstructed *en-face* plots for a representative lumbar FCL surface using reflectivity (Figure 5A), and maximum power (Figure 5B). Integrating the depth information from the PS-OCT reflectivity data created the *en-face* reflectivity image, which is comparable to a traditional light microscopy image. The *en-face* images of maximum power, henceforth called “maximum power maps,” resulted from the power spectral analyses described previously. Both images exhibited similar structural patterns, but the pixel value distribution of the maximum power maps created more texture within the image.

To assess the texture of each type of image, one *en-face* reflectivity image and one maximum power map were each digitally deformed using a transformation matrix in Matlab to 10% and 20% equibiaxial extension and 10% and 20% shear. An in-house image-correlation-based 2D tracking algorithm²⁹ was used to track the displacement of pixels between the original, first deformation, and second deformation for each deformation of each image type. Kernel size was determined by image texture, and smoothness was determined by the element size of the mesh. The same parameters and mesh were applied to all images and deformations. The pixel-wise displacements calculated by the algorithm were compared to the actual deformation and the average RMS displacement error was calculated for each condition.

To assess further motion tracking capabilities of each type of image, three *en-face* reflectivity images and three maximum power maps were generated for each specimen, corresponding to REST, BEND 1 and BEND 2. The same pattern occurred in each surface map from REST to BEND 2 but in a slightly different location due to tissue deformation, allowing the aforementioned image-correlation-based 2D tracking algorithm²⁹ to track the displacement of pixels between the three maps of each type. On average, $52.2 \pm 18.1\%$ of the surface was tracked; remaining locations were incompatible with the 2D tracking algorithm due to insufficient signal quality and were removed from further analysis. Within a sample, the same parameters and mesh were used for both the *en-face* reflectivity images and maximum power maps. The correlation coefficients for each node of the applied meshes ($n = 441$ nodes/mesh) were extracted from the image correlation results to determine the most accurate image set of tissue characteristics to use for further analysis; a perfect correlation of an area in two sequential images has a correlation coefficient of 1. The correlation coefficients of each node were pooled for all samples from REST - BEND 1 and from BEND 1 - BEND 2 for the *en-face* reflectivity images and maximum power maps.

Nodal displacements in 2D were represented using a rectangular finite element mesh with piecewise biquadratic basis functions. Whole-tissue deformation gradient tensors were calculated for each sample by averaging the deformation of each element within the mesh.

In addition, the depth of each point was extracted by obtaining the location of the first index inclusive of signal after low-pass filtering. Thus, for each initial location (X,Y,Z) on the surface of the FCL, the new $x-y$ (*en-face*) position was obtained from the image correlation, and the new z position was obtained from the depth measurement, providing a full 3D displacement field for the surface.

Lagrangian surface strains were given by¹³

$$E_{xx} = \frac{x_x^2 + y_x^2 - 1}{2(1 + Z_x^2)} + \frac{z_x^2 - Z_x^2}{2(1 + Z_x^2)} \quad (3)$$

$$E_{yy} = \frac{x_y^2 + y_y^2 - 1}{2(1 + Z_y^2)} + \frac{z_y^2 + Z_y^2}{2(1 + Z_y^2)} \quad (4)$$

$$E_{xy} = \frac{x_x x_y + y_x y_y - X_x Y_x}{2(1 + Z_x^2)(1 + Z_y^2)} + \frac{z_x z_y + X_x Z_y + Y_x Z_x}{2(1 + Z_x^2)(1 + Z_y^2)} \quad (5)$$

where (X,Y,Z) is the initial position of a point, (x,y,z) is the final position, and x_x is $\partial x / \partial X$, etc. The first term on the right-hand side of each equation corresponds to the strain contributions from in-plane deformation (E^{in}), and the second term corresponds to strain contributions from out-of-plane motion (E^{out}). For each node, a 3×3 node biquadratic element was constructed with the node of interest in the center. The heights and displacement over the element were then fit to biquadratic basis functions, which were used to calculate the derivatives needed for equations (3)–(5). RMS surface strains were calculated for each strain map and for every specimen to compare the strain magnitudes across all specimens.

Surface curvature of the REST, BEND 1 and BEND 2 cases was calculated from the height of the samples and the nodal displacement. For each case, the surface curvature was calculated via the function ‘surfature.m’ downloaded from the Matlab Central File Exchange. The function calculated the first and second derivatives of the height (Z) and the first and second fundamental coefficients from a 2D XY grid and the height at each grid point to output the first and second principal curvatures, the Gaussian curvature, and the mean curvature. The principal curvature (κ) in a direction is defined to be the reciprocal of the best-fit circle to the curve and has units of inverse length. The Gaussian curvature (K) is the product of the principal curvatures calculated via the unit normal vectors to the surface. The Gaussian curvature is an intrinsic property of the surface and determines whether the surface is locally convex or saddle-like. The mean curvature (H) is the average of the principal curvatures and determines a local peak or local valley.

RESULTS

Assessment of Image Quality

The algorithm tracked both equibiaxial and shear deformations for the *en-face* reflectivity and maximum power maps with sub-pixel error from the actual applied deformations. Thus, both the *en-face* reflectivity images and the maximum power maps could be used to extract the biaxial extension and shear deformations applied in this study based on the deformation of a single image. However, an additional assessment was necessary to correlate the REST, BEND 1 and BEND 2 images, which had a similar pattern and texture, but were not identical due to tissue deformation. For the *en-face* reflectivity images, the mean correlation coefficients of every node in each mesh for all five samples from REST - BEND 1 and BEND 1 - BEND 2 were both 0.56 ± 0.13 ($n = 2205$ nodes each). For the maximum power maps, the mean correlation coefficients from REST - BEND 1 and BEND 1 - BEND 2 were both 0.69 ± 0.13 ($n = 2205$ nodes each). The maximum power maps were selected for further analysis because of the significantly higher ($p < 0.0001$, $n = 2205$ nodes) correlation between sequential images determined by the algorithm.

In-Plane Tissue Kinematics

The maximum power maps for the three cases all shared the same general surface pattern, but in different locations. Figure 6 includes maximum power maps for a representative sample for the three cases. Banding of high-signal (yellow) spanned medially-laterally between the articular facets. The high-signal banding was approximately $200 \mu\text{m}$ in width, comparable to the diameter of lumbar FCL collagen fiber bundles³². Low-signal areas corresponded to areas of poor transmission quality from the original PS-OCT scan.

Pixel-wise displacements tracked across the three sequential surface images gave a grid of local displacement vectors with a range of magnitudes and orientations. The overlaid 20×20 element square mesh (Figure 6) varied in size with the usable area of the individual image. The average mesh size was 22.3 mm^2 , and the average element was of 0.0559 mm^2 resolution. Displacement vectors calculated from the motion tracking (Figure 7) were predominantly oriented in the superior-inferior direction, following the gross motion applied by vertebral displacement. Another way to visualize the displacement of the overlaid mesh is by plotting the nodes of the mesh on the surface of the retardance images. The displacement from REST to BEND 2 of the column of nodes denoted by the arrow in Figure 7A is plotted on their corresponding retardance images. The deformed nodes travel laterally and traverse image stacks from the initial position. Because the nodes follow the inherent structural characteristics of the lumbar FCLs, the banding frequency under each node appears similar in corresponding nodes. Samples also exhibited gross shear in response to the applied displacement, as was expected from physiological motion, evidenced by the non-zero diagonal terms of the whole-tissue average deformation gradient tensor

$$F = \begin{bmatrix} 0.99 & 0.01 \\ -0.03 & 1.03 \end{bmatrix} \text{ across all samples.}$$

Each sample had its own unique and complex strain field, including normal strains as well as shear, but some broad trends emerged as shown for a representative sample (Figure 8)

and across all samples ($n = 5$) in Figure 9. In-plane strain fields of a representative sample are shown in Figure 8A. The average RMS strain in the off-fiber direction was higher ($E_{YY}^{in}=0.049$) than the fiber direction strains ($E_{XX}^{in}=0.031$) across all samples, although not significant ($p > 0.05$, $n = 5$). This result is as expected, because the medially-laterally oriented collagen fibers are stiffer along their long axis, and were less stretched by the imposed motion. The averaged RMS in-plane strain contribution of the shear component ($E_{XY}^{in}=0.035$) was comparable to the normal strains, consistent with the gross appearance of shearing.

Out-of-Plane Tissue Kinematics

Sample height increased in all samples with bending. The medial border of the lumbar FCLs was consistently the apex of the surface, which sloped downward toward the lateral, superior and inferior edges. The average mean curvature (H) of all samples increased in magnitude from $-0.0392 \pm 0.1592 \text{ mm}^{-1}$ to $-0.0540 \pm 0.1301 \text{ mm}^{-1}$ with bending of the facet joint ($p > 0.1$ by a two-tailed paired t-test, $n = 5$). Local mean curvature (H) is plotted on the changing surface of a representative FCL in Figure 10. The p-value and large standard deviations indicate there was a large variation in curvature between samples, but the curvature within each individual sample increased consistently. All samples exhibited local peaks and valleys aligned in the medial-lateral (fiber) direction along the long axis of the collagen fibers. Local decreases in curvature are found in the superior half of all samples, whereas the inferior half of the sample tends to increase in curvature.

Sample out-of-plane deformation (Figure 8B) created strains (E^{out}) roughly an order of magnitude smaller than strains resulting from the in-plane deformation (E^{in}). The average RMS out-of-plane strain contribution (E^{out}) magnitude was significantly higher in the off-fiber direction ($E_{YY}^{out}=0.014$) than along the fibers ($E_{XX}^{out}=0.0032$) ($p < 0.020$, $n = 5$) and followed the same trend shown for in-plane strains (E^{in}). The superior-inferior term E_{YY}^{out} varied between positive (the in-plane strain calculation underestimated the amount of extension because the tissue surface was steepening) and negative (in-plane calculation overestimated extension because the tissue surface was flattening). This variation had a banded appearance that followed the curvature bands of Figure 10, as expected.

Three-Dimensional Surface Strains

Three-dimensional surface strains (E) are a summation of the local in-plane strain (E^{in}) and out-of-plane strains (E^{out}) (Figure 8C). Similar in magnitude to E^{in} , the 3D average RMS strain magnitude in the off-fiber direction was higher ($E_{YY} = 0.052$) than fiber axis strains ($E_{XX} = 0.032$) across all samples, although not significant ($p > 0.05$, $n = 5$). The averaged shear component of the 3D RMS strain contribution ($E_{XY} = 0.038$) was again quite large, comparable to the in-plane strains.

In-Plane vs. Three-Dimensional Surface Strain Comparison

Two-dimensional planar strains (E^{in}) are excellent predictors of the true three-dimensional strain (E). To understand the connection between E^{in} and E , local points from all samples were selected that experienced no change in mean curvature (H) with bending, defined by a

change in curvature less than 1%, ($|H_{BEND2} - H_{REST}| < 0.01$). These local points were sorted into two categories, increasing surface slope and decreasing surface slope between REST and BEND 2, based on the change in the unit z-component of the normal vector (n_z), a measure of the surface rotation. As stated previously, increasing and decreasing slopes correspond to $E^{out} < 0$ and $E^{out} > 0$, respectively. Linear regression was used to determine the line of best fit; results are depicted in Figure 11. E_{xx}^{in} are equivalent to E_{xx} with increasing slope (Figure 11A); E_{xx}^{out} are negligible likely because of restriction by the stiff collagen fibers meaning the tissue does not tilt along the lateral to medial fiber axis. However, E_{xx}^{in} underestimates E_{xx} with decreasing surface slope (Figure 11B), showing that the tissue surface only becomes more flat in the lateral to medial fiber axis. This is expected because facet joint motion creates shear across the FCL and elongates the off-fiber axis. E_{yy}^{in} overestimates E_{yy} with increasing slope (Figure 11C) and E_{yy}^{in} underestimates E_{yy} when the tissue is flattening (Figure 11D). Evidence of this trend is witnessed pictorially in Figure 8.

DISCUSSION

Polarization-sensitive optical coherence tomography (PS OCT) is non-destructive, reflection-based imaging modality that has been shown^{2, 9, 14} to provide depth-resolved structural characteristics of biological tissues. The current study shows the use of phase retardance images produced by polarization-sensitive optical coherence tomography (PS OCT) for providing the means for marker-free, high-resolution 3D motion tracking of the lumbar facet capsular ligament (FCL). Maximum power maps, extracted from the inherent fiber alignment of the FCL, depict the medially-laterally orientated collagen fiber bundles of the lumbar FCL as bands of high signal. The distinct features of the maximum power maps were passed through an image-correlation-based algorithm²⁹ to track surface motion in lieu of standard fiducial markers. The in-plane strains from the image correlation, combined with out-of-plane deformation (provided by the depth resolution of the system), allowed calculation of the three-dimensional Lagrangian surface strains. Thus, PS OCT can measure high-resolution structural and kinematic information about biological tissues within the imaging depth by non-contact means. The current study utilized the deforming orientation state of the tissue only as a fiducial marker to track 3D surface motion, but the orientation state could also be used for its inherent value, such as distinguishing fiber alignment in tissue-engineered tendon¹ following fiber pathways in the brain³⁶.

The study of motion via OCT has been previously reported in other contexts. OCT was used to identify gross strain and strain rate based on heat wall thickness deformation by Li et al.²⁰ and 3D surface motion of the embryonic chick heart was monitored by Filas et al.^{13, 14} using microspheres as markers. A major feature of the current technique is the use of intrinsic tissue features to track motion, rather than the use of external markers or measurements. It is also, to our knowledge, the first application of PS OCT to track motion in a musculoskeletal setting.

Other imaging modalities are commonly used for non-destructive motion tracking, such as ultrasound imaging^{21, 25} and magnetic resonance elastography (MRE)^{23, 5, 15}. Ultrasound

imaging has the capacity to reach much greater penetration depths than PS OCT, but there exists a trade-off between depth penetration and image resolution. In addition, the ultrasound probe must be in direct contact with the tissue of interest, as exposure to air will inhibit image acquisition. MRE combines magnetic resonance images with shear waves to create an elastogram capable of measuring the stiffness of soft tissues. However, MRE is expensive, bulky, and would not be feasible in an operating room setting because of metal surgical tools and tables. Therefore, there is room for a technique that could monitor surface motion and mechanical properties in high-resolution during surgery. Because PS OCT is a non-contact and depth-resolved technique, it could serve as the basis for a new device for imaging spinal tissues or other tissues in surgery.

Development of the current method is motivated by the high prevalence of low back pain⁴ and facet joint pain, which is often misdiagnosed and mistreated due to lack of understanding¹⁰. The FCL is densely innervated¹⁷ with mechanosensitive nociceptive and proprioceptive nerve fibers that are embedded within the collagenous fiber bundles. Therefore, the ability to track the 3D kinematics of the FCL and extract the surface strains using 3D structural data may elucidate how these nerve fibers selectively activate in order to transmit either positional information or the perception of pain. Additionally, PS OCT scans could quantify abnormal strain patterns or alteration of fiber arrangement within the FCL or another tissue of interest.

A limitation of the current method is it is unable to characterize the deeper portions of the lumbar FCL. The lumbar FCL is approximately 3 mm thick, and the current PS OCT system can only penetrate up to 1 mm of tissue depth. However, the imaging range of PS OCT can be increased by application of tissue clearing agents^{33, 31}. Although the phase retardance may be lowered by clearing, the slope of the phase retardance can still be used for analysis of the cleared tissue. Clearing the tissue to obtain a greater depth resolution may still allow motion tracking as shown here. Moreover, PS OCT is advantageous over transmission-based modalities, such as polarized light imaging (PLI), which are incapable of imaging a tissue of this thickness at all and which would be impossible to use in an intact (living or cadaveric) spine because of the surrounding tissues.

In conclusion, we have demonstrated that the use of phase retardance images produced by PS OCT provides the means for 3D motion tracking of curved surfaces without the use of external fiducial markers. The method shows that surface motion can be tracked by utilizing inherent structural properties of the tissue. This approach could be combined with methods such as those of Boyle et al.⁶ or Witzenburg et al.³⁸ to identify damaged or remodeled regions in a tissue.

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Figure 1.

Schematic of preparation of an L3–L5 motion segment for imaging. **A.** The posterior view of the motion segment shows the bilateral lumbar FCLs, which flank the central spinous processes. All posterior soft tissue was removed except the FCLs and ligamenta flava. **B.** The right lateral view shows the anterior elements, which include the vertebral bodies and intervertebral discs (IVDs). The anterior elements were removed at the pedicles and bone screws were inserted into L3, L4 and L5 laminae of the motion segment. Spinous processes were also removed to create a larger window for FCL imaging. **C.** The same right lateral view after removal of anterior elements, spinous processes and insertion of bone screws into the laminae. **D.** A representative specimen after preparation.

TOP

Figure 2.

The L3–L5 motion segment bends when bone screws are squeezed. **A.** The L5 articular facets are fixed within a force grip and towel forceps are placed around the L3 and L5 bone screws. **B.** Schematic of resting orientation. **C.** The forceps squeeze the bone screws together, which bends the motion segment at both the L3–L4 and L4–L5 facet joints. **D.** Schematic of a bent motion segment. Bone screw displacement (x) and corresponding L4–L5 facet joint rotation angle (θ_R) are recorded.

TOP

Figure 3.

Specimen placement for imaging by OCT. **A.** The specimen was placed on a custom stage, which is securely bolted to the tabletop. The top of the stage had a thin slit to allow the motion-inducing bone screws to protrude into the stage cavity and a side window provided access for the forceps. **B.** A top-down view shows a right FCL aligned under the OCT laser. Spinous processes were removed so the beam could be focused closer to the FCL without interfering with the specimen. **C.** Image acquisition began at the lateral FCL. Each frame was acquired in the superior-inferior direction, and the frames were stacked from the lateral

aspect to the medial aspect. Polymer retarder film with a jagged edge was included within the frame to enable image registration.

TOP

Figure 4.

Image processing flow diagram of a representative. First, the image stack, composed of retardance maps, is filtered in 3D with an ideal low-pass filter. A retardance map includes hundreds of vertical a-lines that can be represented simultaneously as (A.) an image, or (B.) the signal within an individual a-line can be plotted. Note how the color bar corresponds to the wave pattern. The a-line is transformed into Fourier space and (C.) a power spectral analysis found the dominant banding frequency (x-axis value) and the maximum power (y-axis value) of the signal. D. The dominant banding frequency and the maximum power of

the a-line are assigned a color based on its value and plotted with remaining a-lines as a 2D array.

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Figure 5.

Reconstructed *en-face* (A.) reflectivity and (B.) maximum power images for a representative lumbar FCL surface. **A.** The *en-face* reflectivity image was calculated by integrating the depth information from PS OCT and is similar to a conventional image one would expect from a microscope. **B.** The *en-face* maximum power image plots the maximum power from the spectral analysis. Both plots were created to determine which type of inherent tissue characteristic is best to use in an effort to track the in-plane motion of the FCL. The most desirable plot type is one that has a reproducible, abundant and distinct texture. The texture in both plots was analyzed with an image-correlation-based tracking algorithm²⁹ to assess the displacement error from an applied known deformation and to assess the correlation between REST, BEND1 and BEND 2 images for each tissue characteristic.

TOP

Figure 6.

Representative maximum power maps overlaid with a 20x20 element mesh. The three maximum power maps have the same general pattern, but the location of the pattern changes slightly. The REST nodes remained linked to their initial features, thus the nodes displaced as the locations of the features shifted in BEND 1 and again in BEND 2. Nodal displacements were used to calculate in-plane surface strains.

Figure 7.

Nodal displacements from Figure 6 are (A.) represented as sequential displacement vectors, and (B.) plotted on the surface of retardance maps. **A.** Displacement vectors show the surface primarily moved in the superior-inferior direction. Black dots show REST nodes, maroon arrows show their displacement from REST to BEND 1, and orange arrows show the displacement from BEND 1 to BEND 2. **B.** A column of nodes denoted by the black arrow in A. is plotted on the surface of their corresponding retardance maps for REST and BEND 2. Nodal displacements follow the features of the maximum power maps, and

maximum power is related to retardance. Thus, the banding pattern under each node is similar between corresponding nodes in REST and BEND 2 because the 2D surface tracking is dependent on the internal structure.

TOP

Figure 8.

Representative strain fields from REST to BEND 2 depict (**Row A.**) the in-plane strains (E^{in}) and (**Row B.**) the out-of-plane strains (E^{out}), which sum to (**Row C.**) the 3D surface strains (E). Strain fields are plotted on the 2D deformed surface of BEND 2 and show significant shear. The zero contour line is shown and the color bar applies to all plots. **Row A.** In-plane strains (E^{in}) are derived from 2D displacement vector fields shown in Figure 7. The large strains in the inferior-medial corner of E_{YY}^{in} correspond to the large inferiorly oriented displacement vectors shown in the same region of Figure 7. E_{YY}^{in} are predominantly

the largest of E^{in} , whereas about half of E_{xx}^{in} are compressive. **Row B.** Out-of-plane strains (E^{out}) are a product of surface curvature changes and surface rotations. E^{out} are an order of magnitude smaller than E^{in} . When $E^{out} < 0$, the surface increases in slope and when $E^{out} > 0$, the surface is flattening. The majority of E_{xx}^{out} are greater than zero, meaning the surface along the fiber axis. Negative regions of E_{yy}^{out} exist and are aligned with the fiber axis; these surface areas rotate to increase in slope while the remainder of the sample becomes increasingly flat. **Row C.** Overall, 3D surface strains (E) are similar in magnitude to E^{in} , meaning E^{in} are good predictors of E . Increases in E are visible where both E^{in} and E^{out} are positive, and E decreases where E^{out} are compressive, such as the superior edge of E_{yy} compared to the superior edge E_{yy}^{in} .

TOP

Figure 9.

RMS surface strains for all samples ($n=5$). The median value for each type of RMS strain is displayed with 95% confidence intervals. In following with the representative sample in Figure 8, E_{YY}^{in} and E_{YY} for all samples have the largest RMS strains and additionally, the largest range of values. E_{YY}^{out} are significantly larger than E_{XX}^{in} and E_{XY}^{in} ($p < 0.05$, $n = 5$). These large strains, indicative of greater tissue motion, are found in the off-fiber axis, which is less stiff than the fiber axis. E_{XX}^{in} and E_{XX} are not significantly different ($p > 0.05$, $n = 5$).

from another because these strains are calculated in the stiff fiber direction and very little E_{xx}^{out} is accumulated from the increase in curvature.

TOP

Figure 10.

The shape of the FCL surface changes with bending as depicted by the overlaid mean curvature (H). A value of $H > 0$ denotes a local valley and a value of $H < 0$ signifies a local peak. The inferior-lateral corner decreases in height and in curvature from REST to BEND 2. This follows with the high inferior-lateral E_{YY}^{in} in Figure 8A and the large displacement vectors in Figure 7 in the same region.

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Figure 11.

In-plane strains E^{in} are excellent predictors of 3D surface strains (E). Local points that exhibited no change in curvature were gathered for all samples and sorted into two groups, increasing slope and decreasing slope based on negative or positive values of E^{out} , respectively. **A.** E_{XX}^{in} are equivalent to E_{XX} because the stiff collagen fibers restrict the surface slope from increasing with induced bending. **B.** Conversely, E_{XX}^{in} underestimates E_{XX} as the surface flattens, following the shear response created by rigid facet joint motion. Perpendicular to the fiber axis, **(C.)** E^{in} overestimates E_{YY} when the surface slants and **(D.)**

underestimates E_{YY} when the surface flattens. This over and under prediction is as expected due to the nature of the rotations.