Imaging myocardial fiber orientation using polarization sensitive optical coherence tomography

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Abstract: Knowledge of myocardial fiber architecture is essential towards understanding heart functions. We demonstrated in this study a method to map cardiac muscle structure using the local optical axis obtained from polarization-sensitive optical coherence tomography (PSOCT). An algorithm was developed to extract the true local depth-resolved optical axis, retardance, and diattenuation from conventional round-trip results obtained in a Jones matrix-based PSOCT system. This method was applied to image the myocardial fiber orientation in a bovine heart muscle sample.

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OCIS codes: (110.4500) Optical coherence tomography; (230.5440) Polarization-selective devices.

References and links

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1. Introduction

Fiber organization in cardiac muscle plays an important role in heart functions such as electrical potential conduction [1] and force production [2,3]. In addition to traditional histology image analysis [4], diffusion-tensor magnetic resonance imaging (MRI) [5] has been intensively investigated for visualizing the cardiac muscle fiber organization. Ultrasound elastic tensor imaging [6] has also been explored due to its faster acquisition speed. In addition, optical coherence tomography (OCT) has been explored to quantify fiber orientation in heart muscles [7,8]. Although OCT has a smaller field of view and penetration depth than MRI and ultrasound imaging, its superior spatial resolution makes it possible to reveal fiber organization at the microscopic level. However, previous OCT studies rely on the intrinsic intensity contrast in OCT structure images, ultrahigh resolution and advanced image processing techniques [9] are necessary to identify fiber bundles.

Fibrous structure is associated with optical birefringence which can be characterized by using optical polarization imaging. A few studies have applied polarization-sensitive optical coherence tomography (PSOCT) for imaging fiber orientations in various fibrous tissues. Nakaji et al. [10] determined nerve fiber orientation by analyzing the polarization direction of the backscattered light when rotating the polarization direction of a linearly polarized incident light. The polarization direction of the backscattered light is the same as the incident light when the incident polarization is aligned with the tissue fiber orientation. PSOCT was also applied in a multi-contrast OCT system to measure the differential retardance which was then used along with relative optical axis to indirectly reconstruct neural fiber pathways [11]. In fact, optical axis orientation can be directly imaged in PSOCT. However, due to the round-trip measurement in conventional PSOCT systems, the directly measured optical axis doesn’t represent the true “local” fiber orientation in samples with depth-varying optical axis [12,13]. We recently developed an algorithm that can obtain local optical axis from the round-trip measurement in a conventional PSOCT system with a single circularly polarized incident light [13]; however, this method is only applicable for samples with negligible diattenuation.

Here we report an improved algorithm to extract the complete set of local polarization properties including local retardance, local diattenuation, and local axis orientation using a Jones matrix based PSOCT [14]. The obtained local optical axis was then applied to image the complex myocardial fiber orientation in a bovine heart sample.

2. Method

The Jones matrix PSOCT system used in this study has been described in detail elsewhere [14]. The Jones matrix measurement was carefully calibrated for imaging sample structure, “relative” retardance, diattenuation and optical axis as shown in [14]. The polarization parameters directly measured in PSOCT are denoted as “relative” because they are affected by the entire sample properties from the surface to the measurement depth. They do not represent the true local tissue properties [12,13] in general. The part of the algorithm for calculating “local” retardance and diattenuation was reported previously by Makita et al. [15]. The algorithm for deriving local optical axis was extended from our previous algorithm [13]. The key enhancement here is that the effect of tissue diattenuation was eliminated in the optical axis calculation. The major steps of this new algorithm are outlined below.

The local polarization properties of the tissue segment covered by a single image pixel include: retardance $\delta$, diattenuation $\sigma$, and fast optical axis $\theta$. Retardance and diattenuation are combined into a complex retardance $\gamma = \delta + i\sigma$ with the assumption that they share the same optical axis. The sample segment at a depth $i$ can be represented using the Jones matrix $J_L(\theta_i, \gamma_i)$ that describes a linear retarder with a complex retardance $\gamma_i$:

$$J_i = J_L(\theta_i, \gamma_i) = R(\theta_i)A(\gamma_i)R(-\theta_i),$$

(1)
where \( \mathbf{R}(\theta_i) \) is the standard rotation matrix \([\cos \theta_i, -\sin \theta_i; \sin \theta_i, \cos \theta_i]\) with \( \theta_i \) being the local fast optical axis. \( \mathbf{A}(\gamma_i) \) is a diagonal complex matrix with complex retardance \( \gamma_i \):

\[
\mathbf{A}(\gamma_i) = \begin{pmatrix}
\cos(\gamma_i/2) & 0 \\
0 & \cos(\gamma_i/2)
\end{pmatrix}
\]

The single-trip Jones matrix from the sample surface (denoted as the 1st pixel) to the \( n \)th pixel in depth is the product of a series of linear retarders, which is equivalent to a general retarder:

\[
\mathbf{J}_{ST}(n) = \mathbf{J}_{L1} \mathbf{J}_{L2} \cdots \mathbf{J}_{Ln} = \mathbf{R}(\alpha_{n}) \mathbf{J}_{L}(\phi_{n}, \rho_{n}) = \mathbf{R}(\alpha_{n}) \mathbf{A}(\rho_{n}) \mathbf{R}(\phi_{n}),
\]

where \( \mathbf{R}(\alpha_{n}) \) is the Jones matrix of a rotator as part of the general retarder. The \( \rho_{n} = \kappa_{n} + i\zeta_{n} \) is the “relative” complex retardance with \( \kappa_{n} \) and \( \zeta_{n} \) being the “relative” conventional retardance and diattenuation, respectively; \( \phi_{n} \) is the “relative” optical axis. The round-trip Jones matrix for the \( n \)th layer that is measured directly in PSOCT can be represented as:

\[
\mathbf{J}_{RT}(n) = \mathbf{J}_{L1} \mathbf{J}_{L2} \cdots \mathbf{J}_{Ln} = \mathbf{R}(\phi_{n}) \mathbf{A}(2\rho_{n}) \mathbf{R}(\phi_{n}) = \mathbf{J}_{L}(\phi_{n}, 2\rho_{n}),
\]

where the rotator matrix \( \mathbf{R}(\alpha_{n}) \) in Eq. (3) is cancelled and the relative complex retardance is doubled due to the round-trip measurement.

Once \( \rho_{n} \) and \( \phi_{n} \) are obtained from \( \mathbf{J}_{RT}(n) \), the Jones matrix \( \mathbf{J}_{L}(\phi_{n}, \rho_{n}) \) in Eq. (3) can be constructed following the Eq. (1). To derive the local complex retardance for the \( (n + 1) \)th layer \( \gamma_{n+1} \), a new Jones matrix is constructed using the measured round-trip Jones matrix at depth of \( (n + 1) \) and \( \mathbf{J}_{L}(\phi_{n}, \rho_{n}) \) at depth \( n \):

\[
\mathbf{J}_{STM}(n) = \mathbf{J}_{STM}^{T}(n-1) \mathbf{J}_{STM}(n) = \mathbf{R}(\phi_{n}) \mathbf{A}(\rho_{n}) \mathbf{R}(\phi_{n}),
\]

According to similar matrix transformation [12,15], the eigenvalues of \( \mathbf{N}^{n+1} \) are identical to those of the local Jones matrix \( \mathbf{J}_{STM}^{n+1} \mathbf{J}_{STM}^{-1} = \mathbf{J}_{L1} \mathbf{J}_{L2} \cdots \mathbf{J}_{Ln} = \mathbf{R}(\phi_{n}) \mathbf{A}(\rho_{n}) \mathbf{R}(\phi_{n}) \).

To obtain the local optical axis, a modified round-trip cumulative Jones matrix \( \mathbf{J}_{RTM}(n) \) was first constructed from the measured \( \mathbf{J}_{RT}(n) \) by removing the diattenuation i.e. the imaginary part of the relative complex retardance \( \rho_{n} \):

\[
\mathbf{J}_{RTM}(n) = \mathbf{R}(\phi_{n}) \mathbf{A}(2\kappa_{n}) \mathbf{R}(\phi_{n}),
\]

where \( \kappa_{n} \) and \( \phi_{n} \) are the measured relative conventional retardance (real number) and axis [14]. The local Jones matrix of \( n \)th layer can be obtained by removing the single-trip Jones matrix of the \( (n-1) \)th layer from \( \mathbf{J}_{RT}(n) \):

\[
\mathbf{J}_{STM}(n) = \mathbf{J}_{STM}^{T}(n-1) \mathbf{J}_{STM}(n) \mathbf{J}_{STM}^{-1}(n-1) \mathbf{J}_{STM}(n),
\]

where \( \mathbf{J}_{STM}(n) \) is the modified single-trip Jones matrix with diattenuation removed. The local optical axis \( \theta_{n} \) can be calculated from \( \mathbf{J}_{STM}(n) \) using eigen-decomposition. This calculation needs to be conducted iteratively from the surface where \( \mathbf{J}_{1} = \mathbf{J}_{STM}(1) \). At each new iteration, the local Jones matrix \( \mathbf{J}_{n} \) was constructed using the obtained local optical axis \( \theta_{n} \) and previously obtained local retardance \( \delta_{n} \) (Eq. (5)) as \( \mathbf{J}_{n} = \mathbf{R}(\theta_{n}) \mathbf{A}(\delta_{n}) \mathbf{R}(\theta_{n}) \). The single-trip Jones matrix for the next iteration was then updated as \( \mathbf{J}_{STM}(n) = \mathbf{J}_{n} \mathbf{J}_{STM}(n-1) \).

Because the fast optical axis is orthogonal to the fiber orientation, the obtained fast optical axis was shifted by 90° to represent the fiber orientation. In other words, the slow optical axis was denoted as “optical axis” in the results to representation the myocardial fiber orientation.
The zero degree was defined as the B-scan direction with counter-clockwise defined as positive direction.

3. Results and discussion

The above processing was applied to visualize the myofiber organization in a piece of bovine cardiac muscle sample. Fresh bovine heart was obtained from the meat science laboratory at the University of Missouri-Columbia. Small pieces of the tissue samples with a dimension of 2cm × 2cm × 1cm in length × width × thickness were dissected from the inner wall of the right ventricle. Without any further sample processing, the PSOCT scanned the fresh sample from the endocardial side over an area of 6mm × 6mm (B-scan × C-scan) at room temperature. The 3D image data volume was 1024 × 2000 × 300 pixels (A-scan × B-scan × C-scan). The system (844nm) had a depth resolution of 6.7 μm in air and lateral resolution of 20µm. Its imaging speed was 52k A-lines/sec., and the entire 3D data acquisition took about 23 sec.

Figure 1 shows the conventional cross-sectional PSOCT images [14] and local polarization images. As expected, “banded” patterns appeared in the images of relative retardance $\kappa$ (Fig. 1(c)) and relative optical axis (Fig. 1(b)). The periodic depth profile in relative retardance is due to the phase wrapping in accumulation retardance with depth. The periodic pattern in relative optical axis is due to the phase wrapping in retardance. The relative diattenuation increased with depth and ranged between 0 and 0.25.

Fig. 1. Example PSOCT images of a bovine heart muscle sample. On the first row are cross sectional images of (a) structure, (b) relative optical axis, (c) relative retardance, and (d) relative diattenuation. On the second row are images of (e) local optical axis, (f) local retardance, and (g) local diattenuation. The scale bar in (a) represents 0.5mm.

The aforementioned “banded” appearance disappeared in the images of local retardance (Fig. 1(f)) and optical axis (Fig. 1(e)). The local retardance and local diattenuation images were relatively homogeneous, suggesting a relatively uniform distribution of cardiac muscle structure. However, the local fiber axis in Fig. 1(e) revealed a divided heart tissue sample with the muscle fiber oriented at roughly zero degree at the right side and $-75^\circ$ at the left side.

Fig. 2. 3D PSOCT images of a bovine heart muscle sample: (a) structure; (b) local optical axis; (c) local retardance; and (d) local diattenuation. The 3D image covers a sample volume of 6.0mm × 6.0mm × 2.03mm (in B × C × A).

Figure 2(a) shows the 3D structural image of the tissue volume. On the image surface, traces of muscle fibers are visible at the left side along the C-scan direction. However, the
fibers on the right side are not as obvious. The corresponding 3D volumes of local polarization images are shown in Figs. 2(b)-2(d). Again, comparing with the optical axis image (Fig. 2(b)), the 3D images of local retardance (Fig. 2(c)) and diattenuation (Fig. 2(d)) are more homogeneous. The local retardance had values of 0.004 rad/µm to 0.008 rad/µm and the local diattenuation ranged from $2.5 \times 10^{-3}$ to $7.6 \times 10^{-3}$. The local optical axis image within the B-C plane essentially divided the sample into two color blocks. Consistent with the appearance in surface structural image, the reddish left block had fiber orientation around $-80^\circ$ to $-70^\circ$. The blue-green part on the right indicated orientation around $-30^\circ$ to $30^\circ$.

Figure 3(a) shows the en face structure image extracted from the depth marked as a dashed line in Fig. 1(a) at about 0.3 mm from the surface. Although the fiber tracts can be visualized at the sample surface in the structural image (Fig. 2(a)), they were not so clear at deeper locations due to reduced image contrast. The fiber orientation at the lower-left portion of Fig. 3(a) was still visible because of the relatively smaller depth at this position from the sample surface. Figure 3(b) shows the conventional cumulated optical axis and Fig. 3(c) shows the local optical axis. Apparently the local axis image provided a better view of the fiber orientation than the conventional relative axis which suffered the image artifact from phase wrapping. To better visualize their fiber orientations, the fiber tracts were represented as streamlines using a set of random seeding position from a predefined grid. Streamlines were traced only when the signal intensity in the structural image is greater than the noise level (40dB). As clearly shown in Fig. 3(d), there are roughly two groups of fiber bundles in the tissue sample. The left part of muscle fibers was oriented at an angle of $\sim-75^\circ$ and the right group was aligned horizontally.

To examine the fiber orientation in detail, three small regions of interest (ROIs) (1.0 mm × 1.0 mm) were selected from different locations in the sample (marked in Fig. 3(a)). The ROI-1 was located in the upper-right of the sample with a wavy fiber organization; ROI-2 was located close to the center at the boundary of two different fiber bundles; ROI-3 was located the left part of the sample with relatively uniform fiber organization. Figures 4(a)-4(c) show the streamlines representation of the myocardial fiber organization within the three ROIs extracted at three depths of 0.3 mm, 0.4 mm, and 0.5 mm from the surface. Although similar
wavy fiber tracts appeared at each depth in Fig. 4(a), subtle changes in fiber organization is evident. In Fig. 4(b), the two fiber tracts with different orientation appeared at small depths. The two tracts merged together with increasing depth and became a smoothly curved single tract at depth of 0.5 mm. The parallel fiber organization in ROI-3 was maintained throughout the 200 µm depth range in Fig. 4(c).

Figure 4(d) further shows the local optical axis extracted at the center of the three ROIs in Figs. 4(a)-4(c). The results were averaged over a small area of 3 pixel × 3 pixel in the \textit{en face} plane (B-C plane). It is known that myocardial fibers have a well-organized architecture across the entire heart [6]. Owing to the limited penetration, the PSOCT system only imaged a very small portion of the tissue across the heart wall, which showed a variety of location-dependent profiles in fiber orientation (Fig. 4(d)). The depth curves obtained in ROI-1 and ROI-2 are similar, but the ROI-2 curve is closer to a straight line with fiber orientation changing linearly from −60° to 0° over a 0.5 mm depth. The ROI-3 curve is flat over the first 0.35 mm depth and then shows a sudden increase.

![Image](image_url)

**Fig. 5.** (a) 3D structure image and (b) streamline representation at 0.4, 0.5, and 0.6 mm depth of another piece of sample excised at a different location of the bovine heart. Also shown are (c) relative and (d) local optical axes calculated from the same sample slice in Fig. 1 without removing the diattenuation.

The detailed fiber architecture clearly varied with measurement location in the heart. Figures 5(a) and 5(b) show the 3D structural image and streamline representation of another piece of muscle sample excised at a different location from the same bovine heart. This sample had a relatively uniform fiber orientation in comparison with the sample shown in Fig. 1-4. It’s important to point out that the removal of the diattenuation is critical in achieving consistent results with good image quality in this heart muscle sample. As shown in Figs. 5(c) and 5(d), both the relative and local optical axes calculated under the assumption of zero diattenuation were much more noisy and irregular than those obtained after removing diattenuation (Fig. 1). The overall useful imaging depth in our current PSOCT results appeared to be ~0.6 mm which is limited by the signal-to-noise in the raw OCT signal.

4. Conclusion

In summary, we demonstrated an optical tractography for imaging myocardial fiber orientation using local depth-resolved optical axis measured from a Jones matrix-based PSOCT system. Jones calculus was applied to extract complete local polarization properties including local retardance, local diattenuation and local optical axis from conventional round-trip PSOCT measurements. The local optical axis orientation was derived using a recursive algorithm after removing the diattenuation. In comparison with our previously reported algorithm [13], the current method can be applied to general biological samples with significant diattenuation. In a bovine heart muscle sample, we showed that the organization of myocardial fibers can be visualized using the local optical axis. Details of myocardial architecture and connectivity can be examined using the streamline representation.

Acknowledgments

This project is supported in part by a NSF grant CBET-0643190.