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Shuliang Jiao and Lihong V. Wang
Optical Imaging Laboratory, Biomedical Engineering Program,
Texas A&M University, 3120 TAMU, College Station, TX 77843-3120
Email: LWang@tamu.edu; URL: http://oilab.tamu.edu

ABSTRACT

A multiple-channel polarization-sensitive system based on optical coherence tomography was built to measure the Mueller matrix of scattering biological tissue. The Jones matrix of a sample can be determined with a single scan and subsequently converted into an equivalent non-depolarizing Mueller matrix. As a result, the system can be used to measure the Mueller matrix of soft tissue. Birefringence, axis orientation, and diattenuation can be extracted.
Keywords: Ellipsometry and polarimetry; Optical coherence tomography.

1. INTRODUCTION

Polarization-sensitive optical coherence tomography (PS-OCT) can reveal important polarization information about biological tissue, which is unavailable in conventional OCT.[1–3] The best way to ascertain the optical polarization properties of a sample is to measure its Mueller matrix. The combination of polarimetry and OCT makes it possible to acquire the Mueller matrix of a sample with OCT resolution.[4,5] Jiao *et al*[5] demonstrated that the degree of polarization of the back-scattered light measured with OCT remains unity—indicating that the measured Mueller matrix is non-depolarizing. This conclusion allows the use of the Jones matrix in OCT. The relatively time-consuming process of the previous measurements of Mueller matrices limited the technique to use with stable samples such as bones. To measure unstable samples such as soft tissues, we have developed a system that can determine the Mueller matrix with a single depth scan (A scan). The Jones matrix **J** can be transformed into an equivalent non-depolarizing Mueller matrix by[6] $\mathbf{M} = \mathbf{U}(\mathbf{J} \otimes \mathbf{J}^*)\mathbf{U}^{-1}$, where $\otimes$ represents the Kronecker tensor product and **U** is the 4×4 Jones–Mueller transformation matrix [1, 0, 0, 1; 1, 0, 0, –1; 0, 1, 1, 0; 0, *i*, –*i*, 0], which is written row by row. The Jones matrix has four complex elements, in which seven real parameters are independent.

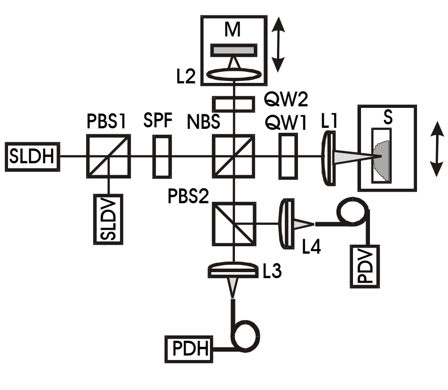

Fig.1 Schematic of the multiple-channel polarization-sensitive OCT system: SLDH and SLDV are superluminescent diodes, horizontally (*H*) and vertically polarized (*V*), respectively; PBS1, PBS2, polarizing beam splitters; SPF, spatial filter assembly; NBS, nonpolarizing beam splitter; QW1, QW2, zero-order quarter-wave plates; M, mirror; L1–L4, lenses; S, sample; PDH and PDV are photodiodes for *H* and *V* polarization components, respectively.

2. METHODS

The schematic of the experimental setup is shown in Fig. 1. Two superluminescent diodes (central wavelength λ = 850 nm, FWHM bandwidth $\Delta\lambda$ = 26 nm) are employed as low-coherence light sources and are amplitude modulated at 3 and 3.5 kHz by variation of the injection current. The two source beams are merged by a polarizing beam splitter, filtered by a spatial filter assembly, and then split into the reference and sample arms by a nonpolarizing beam splitter. The sample beam passes through a quarter-wave plate (λ/4 plate) oriented at 45°, and is focused into the sample by an objective lens (L1: f = 15 mm and NA = 0.15). The reference arm consists of a λ/4 plate oriented at 22.5°, a lens (L2), and a mirror. After double passing through the λ/4 plate, the horizontal polarization (H) of the incident light is converted into 45° polarization, and the vertical polarization (V) of the incident light is converted into −45° polarization. The reference beam combines with the back-scattered sample beam through the nonpolarizing beam splitter. The combined light is split into two orthogonal polarization components by polarizing beam splitter PBS2. The two components are coupled into two single-mode optical fibers, which are connected to photodiodes PDH and PDV, respectively. A data-acquisition board sampling at 50,000/s digitizes the two signals. A Doppler frequency of ~1.2 kHz is generated by the linear scan of the reference mirror. The carrier frequencies are the beat and harmonic frequencies between this Doppler frequency and the modulation frequencies of the light sources.

The incident Jones vectors, $\mathbf{E}_i$, to the sample arm is transformed to the detected Jones vector, $\mathbf{E}_o$, as follows:

$$\mathbf{E}_o = \mathbf{J}_{NBS}\mathbf{J}_{QB}\mathbf{J}_{SB}\mathbf{J}_M\mathbf{J}_{SI}\mathbf{J}_{QI}\mathbf{E}_i = \mathbf{J}_{NBS}\mathbf{J}_{QB}\mathbf{J}\mathbf{J}_{QI}\mathbf{E}_i = \mathbf{J}_T\mathbf{E}_i \tag{1}$$

where $\mathbf{J}_{QI}$ and $\mathbf{J}_{QB}$ are the Jones matrices of the λ/4 plate for the incident and the backscattered light, respectively; $\mathbf{J}_{SI}$ and $\mathbf{J}_{SB}$ are the Jones matrices of the sample for the incident and the backscattered light, respectively; $\mathbf{J}_M$ is the Jones matrix of single backscattering—the same as the one for a mirror; $\mathbf{J}_{NBS}$ is the Jones matrix of the reflecting surface of the non-polarizing beam splitter; $\mathbf{J}$ is the combined round-trip Jones matrix of the scattering media; and $\mathbf{J}_T$ is the overall round-trip Jones matrix. As a result of the reciprocal constraint[7], the matrices $\mathbf{J}$ and $\mathbf{J}_T$ are transpose symmetric if the OCT signal is primarily from single backscattering and the contribution from multiple scattering can be neglected.

We band-pass filter the interference signals with central frequencies of 4.2 kHz and 4.7 kHz and a bandwidth of 10 Hz—the harmonic frequencies of the interference signals of source H and source V, respectively—to extract the interference components of each light source. The two orthogonal interference components (H and V) form the imaginary parts of the output Jones vectors, whose real parts are obtained through inverse Hilbert transformation.[8] When the output Jones vectors for the two independent incident polarization states are determined, the elements of the Jones matrix can then be calculated based on Eq. (1).

3. RESULTS

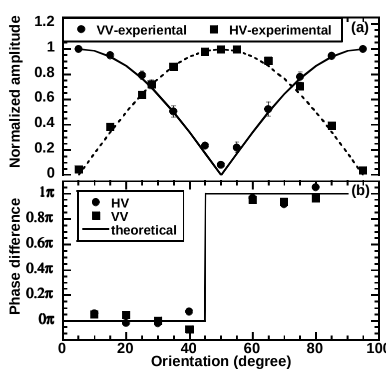

Fig. 2 (a) Normalized amplitude of the vertical components of the measured Jones vectors of a λ/4 plate versus the orientation of the fast axis, HV: horizontally polarized incident light; VV: vertically polarized incident light. (b) Phase differences between the vertical and the horizontal components of the same λ/4 plate. The standard deviations are smaller than the symbols.

The system was first tested by measurement of the matrix of a standard sample—a λ/4 plate at various orientations in combination with a mirror. Figure 2(a) shows the amplitude of the vertical components of the measured Jones vector versus the orientation of the wave-plate, where the amplitude of each Jones vector was normalized to unity. Figure 2(b) shows the phase differences between the vertical and the horizontal components of the Jones vectors. The results show that the measured data agree very well with the theoretical values.

The system was then applied to image soft tissue (a piece of porcine tendon). The tendon was mounted in a cuvette filled with saline solution. The sample was transversely scanned with a step size of 5 μm, and multiple A-scan images were taken. For each A scan, the pixels were formed by averaging the calculated elements of the Jones matrix over segments of 1000 points—the resolution of the system. Two-dimensional (2D) images were formed from these A-scan images and then median filtered. The final 2D images are shown in Fig. 3.

Clear band structures can be seen in some of the images. There is no such band structure present in the M_{00} image, which is the image based on the intensity of the back-scattered light. In other words, the M_{00} image is free of the effect of polarization. We believe that the band structure is generated by the birefringence of the collagen fibers in the porcine tendon.

Although all the polarization properties of a sample are contained in the Mueller matrix implicitly, explicit polarization parameters can be extracted from the matrix. The nondepolarizing Mueller matrix $\mathbf{M}$ can be decomposed by polar decomposition[9]:

$$\mathbf{M} = \mathbf{M}_P \mathbf{M}_R , \quad (2)$$

where $\mathbf{M}_P$ and $\mathbf{M}_R$ are the Mueller matrices of a diattenuator and an elliptical retarder, respectively. In biological tissues, it is reasonable to believe that the orientations of the diattenuator and retarder are the same. In this case, $\mathbf{M}$ is homogenous in the polarization sense and the order of $\mathbf{M}_P$ and $\mathbf{M}_R$ in Eq. (2) is reversible. Because the effect of non-Faraday circular birefringence is canceled in the round-trip OCT signals, only linear birefringence exists in $\mathbf{M}$.

We extracted polarization information from a piece of porcine tendon set at various orientations. The rotation axis of the sample is collinear with the optical axis (direction of incidence). The measurements were made at five different orientations with an interval of 10°. The M_{31} and M_{32} elements in Eq. (2) for a Mueller matrix that contains linear birefringence and linear or circular diattenuation can be expressed as

$$M_{31} = P(P_1, P_2)\sin(2\theta)\sin(\delta) \text{ and } M_{32} = -P(P_1, P_2)\cos(2\theta)\sin(\delta), \quad (3)$$

where P is a function of the principal coefficients of the amplitude transmission P_1 and P_2 for the two orthogonal polarization eigenstates of $\mathbf{M}_P$; θ and δ are the orientation of the fast axis and the phase retardation of the retarder, respectively.

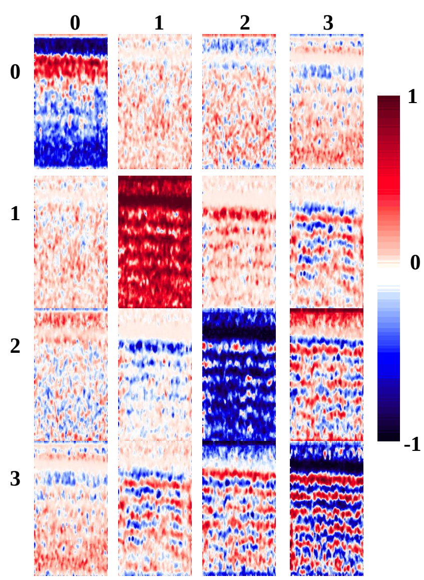

Fig. 3 2D Mueller-matrix (**M**) images of a piece of porcine tendon. Each image except M_{00} is pixel-wise normalized with the M_{00} element and shares the same color table. The size of each image is 0.5 mm × 1 mm.

To increase the signal-to-noise ratio, we laterally averaged every 20 adjacent A scans of M_{31} and M_{32} in the calculated 2D images and fitted them for a physical depth of 0.4 mm from the surface. The calculated birefringence from the fitted data is $(4.2 \pm 0.3) \times 10^{-3}$, which is comparable with the previously reported value of $(3.7 \pm 0.4) \times 10^{-3}$ for bovine tendon.[1] The calculated orientations of the fast axis are (0 ± 4)°, (9 ± 2.9)°, (20.9±1.9)°, (30±2.8)°, and (38±4.3)° after subtraction of an offset angle. The small angular offset is due to the discrepancy between the actual and the observed fiber orientations. The results are very good considering that the tendon was slightly deformed when it was mounted in the cuvette and the rotation axis for the sample may not have been exactly collinear with the optical axis. The diattenuation, defined as $D = (P_1^2 - P_2^2)/(P_1^2 + P_2^2) = \sqrt{M_{01}^2 + M_{02}^2 + M_{03}^2}/M_{00}$, was averaged over all the orientations and linearly fitted over a depth of 0.3 mm. The fitted value of D versus the round-trip physical path length increases with a slope of 0.26/mm and reaches 0.075±0.024 at a depth of 0.3 mm after subtraction of an offset at the surface.

CONCLUSION

In summary, we have developed a novel multiple-channel polarization-sensitive OCT imaging technique. The Jones matrix of a sample can be determined with a single scan. This technique permits the acquisition of 2D tomographic Mueller-matrix images of either hard or soft biological tissues. The Mueller matrix can be decomposed to extract important information on the optical polarization properties of a sample. The polarization properties can potentially be correlated with the conditions of biological tissues and thus used for disease diagnosis.

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REFERENCES

1. J. F. de Boer, T. E. Milner, M. J. C. van Gemert and J. S. Nelson, Opt. Lett. **22**, 934 (1997).
2. M. J. Everett, K. Schoenerberger, B. W. Colston, Jr., L. B. Da Silva, Opt. Lett. **23**, 228 (1998).
3. J. F. de Boer, T. E. Milner, and J. S. Nelson, Opt. Lett. **24**, 300 (1999).
4. G. Yao and L.-H. V. Wang, Opt. Lett. **24**, 537 (1999).
5. S. Jiao, G. Yao and L.-H. V. Wang, Appl. Opt. **39**, 6318 (2000).
6. F. Le Roy-Brehonnet and B. Le Jeune, Prog. Quant. Electr. **21**, 109 (1997).
7. N. Vansteenkiste, P. Vignolo and A. Aspect, J. Opt. Soc. Am. A **10**, 2240 (1993).
8. Y. Zhao, Z. Chen, C. Saxer, S. Xiang, J. F. de Boer and J. S. Nelson, Opt. Lett. **25**, 114 (2000).
9. J. J. Gil and E. Bernabeu, Optik **76**, 67 (1987).