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Authors

Saxer, CE
de Boer, JF
Park, BH
[et al.](#)

Publication Date

2000

DOI

10.1117/12.384176

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Peer reviewed

Fiber-based polarization-sensitive optical coherence tomography

Christopher E. Saxer, Johannes F. de Boer, B. Hyle Park, Yonghua Zhao,

Zhongping Chen, J. Stuart Nelson

Beckman Laser Institute and Medical Clinic, Univ. of California, Irvine

ABSTRACT

In optical coherence tomography (OCT), mapping the polarization state of the reflected light provides additional information about tissue structure and prevents polarization induced image artifacts. As OCT is increasingly used with subjects *in vivo*, demands on the imaging system and data acquisition rates increase. We present a fiber-based, rapid scanning, polarization-sensitive OCT system capable of acquiring image data at the rate of 20K pixels/s. To achieve high scan rates, a rapid-scanning optical delay (RSOD) line generates a reference arm group delay while a waveguide-based phase modulator generates a suitable, stable carrier frequency for the detection electronics. Group delays up to 4.5 mm are obtained at frequencies approaching 1 kHz. The group delay and carrier frequency are independently controllable, which has the advantage that either the lateral or the axial scan direction may be chosen as the fast axis. Tomographic images corresponding to the intensity and polarization components of the Stokes vector describing the backscattered light are obtained by analyzing the interference signals from two orthogonally polarized channels in the detection arm for each of four polarization states incident on the sample. Polarization images of human skin taken *in vivo* are shown.

Keywords: Optical coherence tomography, polarization, fiber, interferometry, Stokes vector, Poincaré sphere

1. INTRODUCTION

Optical Coherence Tomography (OCT) is an emerging technology for non-invasive imaging in biological tissue. Based on a Michelson interferometer, the technique measures spatially resolved reflected intensity in tissue, offering a dynamic range in excess of 100 dB¹ and a recently demonstrated resolution of 1–3 micron.² The development of Polarization-Sensitive OCT (PSOCT) takes advantage of the additional polarization information carried by the reflected light.^{3–7} The two mechanisms that dominate the changes in the polarization state of light propagating through biological tissue are scattering and birefringence. Scattering changes the polarization of light mainly in a random manner. Birefringence alters the polarization state in a prescribed manner dependent on the incident state of polarization and the orientation of the optic axis of the material. Many biological tissues such as tendons, muscle, nerve, bone, cartilage and teeth exhibit linear birefringence due to their linear or fibrous structure. The advantage of PSOCT is the enhanced contrast and specificity in identifying structures in OCT images by detecting induced changes in the polarization state of light reflected from the sample. Moreover, changes in the birefringence may, for instance, indicate changes in functionality, structure or viability of tissues.

2. THEORY

Optical coherence tomography relies on the phenomenon that interference fringes for low-temporal coherence light sources are highly localized about the zero optical path difference (OPD) position of a two beam interferometer. The length of the reference arm may therefore be varied to perform depth ranging within a sample, the interference fringe signal occurring only for light in the sample arm reflected or scattered back from locations within the source coherence length of the zero OPD position. However, in addition to the temporal coherence of the interfering beams, the visibility of the interference fringes depends on the degree to which the polarization of the sample and reference beams match. If, for instance, the polarization of the sample beam, initially the same as that of the reference arm, changes to the orthogonal polarization, no interference signal will be recorded.

The Stokes vector offers a convenient and rather intuitive method to characterize the polarization state of light. The Stokes vector has four components I , Q , U and V where I equals the irradiance of the beam and Q , U and V characterize its state of polarization. An arbitrary, yet hopefully convenient, coordinate system for the plane normal to the direction of propagation must be chosen to define the horizontal (or parallel) and vertical (or perpendicular) directions. The component Q then represents the portion of light linearly polarized along the horizontal ($Q = +1$) or vertical axes ($Q = -1$), U describes the amount of light linearly polarized along $+45^\circ$ or -45° degrees and V represents the amount of right ($V = +1$) or left ($V = -1$) circularly polarized light. For fully polarized light the Stokes parameters obey $I^2 = Q^2 + U^2 + V^2$. For completely unpolarized light $Q = U = V = 0$. If the beam is partly polarized $I^2 \geq (Q^2 + U^2 + V^2)$, and the degree of polarization P may be defined as

$$P = \frac{\sqrt{Q^2 + U^2 + V^2}}{I}. \quad (1)$$

The Poincaré sphere, depicted in Fig. 1, offers a means to visualize the polarization components of the Stokes vector, which when normalized by the irradiance I , range from -1 to $+1$. Note that the Stokes vectors for orthogonal polarizations are collinear, but point in opposite directions. Changes in polarization due to birefringence can be modeled as a rotation of the Stokes (polarization) vector about an axis in the QU plane corresponding to the projected optic axis of the material; the rotation angle is equal to the retardance. For instance, the Stokes vector of a horizontally polarized beam ($\{Q, U, V\} = \{1, 0, 0\}$) incident on a quarter-wave plate with slow axis oriented along $+45^\circ$ ($U = +1$) would rotate 90° ($\pi/4$) about $+U$ to ($\{Q, U, V\} = \{0, 0, 1\}$), which corresponds to right circular polarization. Optical activity, on the other hand, is modeled as a rotation of the Stokes vector about the V axis. Birefringence and optical activity alter the polarization state of light as it propagates through the medium. However, the degree of polarization is preserved.

Fig. 1. The Poincaré sphere and Stokes vector representation of polarization states of light. Linear polarizations are represented by vectors in the QU plane. Right and left circular polarizations correspond to $+V$ and $-V$, respectively.

Since rotation about an axis is an orthonormal transformation, it follows that the angle between Stokes vectors of two different polarization states is preserved upon propagating through a medium with no polarization-dependent losses. This also holds true for a series of birefringent regions with randomly oriented optical axes.

In biological samples the direction of the optic axis is generally not known *a priori*, and indeed may vary in different regions of the tissue. Therefore, PSOCT measurements must be performed in a manner that is sensitive to arbitrary orientations of the optic axis. Previous air-spaced PSOCT configurations probed the sample with circularly polarized light ($\{Q, U, V\} = \{0, 0, 1\}$), which by its rotational symmetry detects polarization changes due to birefringence regardless of the orientation of the sample's projected optic axis. (Note that if the Stokes vector of the input state points along $+V$ in Fig. 1, it will rotate away from this direction upon propagation through a birefringent medium regardless of the orientation of the optic axis in the QU

plane.) The challenge with a single-mode fiber based PSOCT configuration is that the polarization state of the light incident on the sample is difficult to control (and measure) due to fiber polarization mode dispersion (PMD) (*i.e.*, birefringence in the fiber due to random stress and imperfections in the circularity of the fiber core) and stress-induced birefringence, which varies with bending and movement of the sample arm fiber. It is, however, assumed that the incident polarization remains constant within the measurement time of a several A -line scans (1 – 5 ms). Hence, single-mode fiber does not allow simple preparation of the sample beam in a circularly polarized state. If the sample is probed with only one (unknown) polarization state, it is possible for the Stokes vector for that polarization to be collinear with the direction of the optic axis; rotation about this axis would leave the polarization state unchanged despite the presence of birefringence in the sample.

An alternate technique, sensitive to arbitrary orientations of the optic axis, is to probe the sample twice with polarization states that are not collinear in the Poincaré sphere. To probe the sample most effectively, incident polarizations are chosen whose Stokes vectors are perpendicular in the Poincaré sphere. Two suitable examples are right circular polarization for the first state and any linear polarization for the second, or parallel linear ($+Q$) for the first and $+45$ degree linear polarization ($+U$) for the second. For two such polarization states, rotation about any axis in the Poincaré sphere will induce a measurable change in the Stokes vector of one or both of the polarizations regardless of any changes from their initial directions due to PMD or stress-birefringence in the fiber. Without significant experimental complication, two additional polarizations, orthogonal to the first two may be used to probe the sample. Since the Stokes vectors of these two additional input states point opposite to the original states, their polarization component images may be averaged with the respective images of the original two states by subtraction. Thus, four polarization states, whose Stokes vectors lie at 90 degree increments on a grand circle in the Poincaré sphere, are used sequentially to perform interlaced A -line (depth) scans in this fiber-based PSOCT system.

3. EXPERIMENTAL APPARATUS

Previous PSOCT systems were air spaced interferometers using bulk optical components that allowed precise control over the polarization state of light in the sample and reference arms.^{3–7} Fiber based interferometers offer a distinct advantage in terms of system alignment and handling, but as mentioned pose design problems due to polarization changes induced in the optical fiber. With respect to polarization properties, two common types of single-mode fiber are available: conventional single-mode fiber and polarization maintaining (PM) fiber. PM fiber has by design a large birefringence with a beat length (full wave retardance) of 2–3 mm. The energy of the modes polarized along the primary axes of the fiber is preserved, but phase information between the modes is lost since after as little as 30 mm of propagation, the accumulated retardance exceeds the coherence length of the light. Single-mode fiber exhibits Polarization Mode Dispersion (PMD) which is caused by randomly oriented linear and circular birefringence due to stress in the fiber and imperfections in the circularity of the fiber core. The optical path difference between orthogonal polarization states is not linear with fiber length as in PM fiber, but proportional to square root of the length, revealing the underlying one-dimensional random walk nature of PMD. We used single mode fiber (Corning SMF-28) with a PMD ≤ 0.5 ps / km^{1/2}, which insures the polarization-dependent optical path difference to be less than 10 μ m for 4 m of fiber. Furthermore, the PMD of the interferometer was minimized by use of short single-mode fiber lengths in the sample (1.7 m) and reference (0.5 m) arms yielding a total measured PMD estimated at less than 2 μ m.

Figure 2 presents a schematic of the single mode fiber-based PSOCT system. A low-coherence source (AFC technologies) with a bandwidth (FWHM) of 65 nm centered at 1310 nm is polarized by a bulk polarizer and coupled back into a single-mode fiber. Quarter- and half-wave plates before the polarizing beamsplitter are oriented to select the polarization state of the source with the largest power (8 mW), while the quarter- and half-wave plates after the polarizer prepare the polarization such that after propagating through a short length of fiber (150 mm), the light emerges with equal electric field components parallel and perpendicular to the optic axis of a bulk electro-optic polarization modulator (New Focus model 4104).

The polarization modulator can transform the input polarization to any state along the grand circle in the QV plane on the Poincaré sphere. A driving function of four, equal $\pi/2$ phase steps produces the four orthogonal Stokes vectors used for the interlaced A -line scans. The light is then coupled through a 2x2 fiber splitter to the sample and reference arms of the interferometer.

Fig. 2. Schematic of the fiber-based PS-OCT system. The scanning lens assembly pictured in Fig. 3 is used in place of the generic handpiece shown.

The sample arm consists of a fiber with collimator and a focusing lens mounted on a motorized linear translation stage (see Fig. 3). In order to have sufficient depth of focus for a 2 mm *A*-line scan, the minimum spot size of the probing beam is 30 μm . Due to PMD, the exact polarization state at the tip of the sample arm fiber is unknown. However, in a lossless fiber with PMD less than the coherence length of the light, the transformations in the Poincaré sphere are orthonormal, preserving the angles between the four Stokes vectors of the four states exiting the polarization modulator. The dots on the grayed grand circle and axes in Fig. 4 indicate one possible realization of the polarization states at the fiber tip.

Fig. 3. Sample arm scanning lens assembly. The depth of focus is roughly 1.5 mm and the minimum spot size 30 μm .

The reference arm must carry out two critical functions: generate a carrier frequency for the interferometric signal and provide a rapidly varying group delay to effect depth scanning. These are accomplished with a phase modulator and a Rapid Scanning Optical Delay line (RSOD),^{9,10} respectively. A polarization controller (General Photonics PolaRite™) follows the 2x2 splitter in the reference arm and is aligned so that for all four input polarization states, a constant amount of light is transmitted through the PM fiber pigtailed phase modulator (JDS Uniphase), which by its structure also linearly polarizes the light. PM fiber with its axis aligned to the linear polarization is also used to couple the light from the phase modulator into the RSOD. The RSOD is run with the spectrum centered on the galvo mirror, thus generating only a group delay and no phase delay. The carrier frequency of the interferometric signal is generated by the phase modulator only. The phase modulator is driven by a stable, sawtooth waveform at 1 MHz, generating a maximum 2π phase shift after double passage. To reduce the length of fiber in the system and thus the PMD, polarization controllers are used in which stress-birefringence is induced by a

rotating clamp mechanism in a short (140 mm) section of fiber, rather than by winding a long section of fiber about three rotating mandrels as is common in many fiber optic setups.

Fig. 4. The dots on the gray grand circle and axes show a possible realization of the polarization states incident on the sample. In the absence of polarization dependent loss, the 90° angle between the Stokes vectors is maintained from the polarization modulator to the fiber tip of the sample arm. The polarizations are numbered as pairs of orthogonal polarizations.

The detection arm consists of a polarization controller, a fiber-pigtailed polarizing beamsplitter and two detectors. The detection arm polarization controller is aligned such that the reference arm light is split equally over both detectors. The photo-electric signals are high pass filtered, amplified, and digitized by a 12-bit dual-channel 10 Msamples/s per channel A/D board (Gage Applied Sciences, Inc.). The system also has a quasi-real time preview mode in which reflectivity images are acquired, processed and displayed roughly every 3 seconds. This mode aids greatly in selecting a region of interest for the actual polarization-sensitive measurement.

4. DATA ANALYSIS

Data processing relies on digital lock-in detection of the sine and cosine components of the detected interference signal at the 1 MHz reference frequency of the phase modulator. The sine and cosine components of $5 \mu\text{m}$ segments in each 2 mm long A -line (depth profile) are processed to obtain the Stokes parameters as described earlier.^{6,8}

$$I = \sin_H^2 + \cos_H^2 + \sin_V^2 + \cos_V^2, \quad (2)$$

$$Q = \sin_H^2 + \cos_H^2 - \sin_V^2 - \cos_V^2, \quad (3)$$

$$U = 2 \sin_H * \sin_V + 2 \cos_H * \cos_V, \quad (4)$$

$$V = 2 \sin_H * \cos_V - 2 \cos_H * \sin_V. \quad (5)$$

I , Q , U and V represent the components of the Stokes vector and $\sin_H$, $\cos_H$ and $\sin_V$, $\cos_V$ are the sine and cosine components of the horizontal and vertical polarization channels, respectively. The thus calculated Stokes parameters describe the polarization state at the polarizing beamsplitter in the detection arm. Modulating the incident light over four polarization states produces 16 images (I , Q , U and V for each input state). However, since the four input states are two pairs of orthogonal polarizations, these 16 images can be averaged as follows: for each pair of orthogonal incident polarizations, the I images are averaged by addition and the Q , U and V images are averaged by subtraction since their polarization components

Fig. 5. PSOCT images of *in vivo* human forearm skin. Dimensions are $2\text{ mm} \times 2\text{ mm}$ with pixel size $10\text{ }\mu\text{m} \times 10\text{ }\mu\text{m}$ except for the reflectivity image (a) which has $5\text{ }\mu\text{m} \times 5\text{ }\mu\text{m}$ pixels. **(a)** Reflectivity image $\bar{I}$ averaged from four scans of different incident polarizations. **(b)** Retardation phase map indicating the minimum amount of retardance to change the incident polarization vector to the polarization state reflected from a given depth. **(c)** Orthogonality image depicting the value of the inner product of the Stokes vectors of the two averaged polarizations ($S_1 \cdot S_2$). The remaining six images show the polarization components of the Stokes vectors of the backscattered light for polarization S_1 : **(d)** Q_1 , **(e)** U_1 , **(f)** V_1 , and polarization S_2 : **(g)** Q_2 , **(h)** U_2 , **(i)** V_2 .

point in opposite directions (see Fig. 4). The resulting eight images (one set of I , Q , U and V per input pair) define two Stokes vectors that are described by length I_1 and I_2 and direction S_1 and S_2 (calculated from the three components Q , U and V).

Although the exact polarization states with which the sample is probed are not known, an image showing phase retardation map of the tissue can be constructed from the vectors S_1 and S_2 . First, the polarization state incident on the sample is calculated from the average states $\bar{S}_1$ and $\bar{S}_2$ reflected from the glass–tissue interface. Next, at each position in the image, a single rotation matrix $\mathbf{R}$ is calculated that simultaneously transforms $\bar{S}_1$ to S_1 and $\bar{S}_2$ to S_2 . Since an exact solution does not always exist, the best solution for $\mathbf{R}$ is determined by calculating the angles ϕ_1 between $\mathbf{R} \cdot \bar{S}_1$ and S_1 and ϕ_2 between $\mathbf{R} \cdot \bar{S}_2$ and S_2 , then minimizing the sum of the angles, weighted by the absolute length of the Stokes vector, $(\phi_1 I_1 + \phi_2 I_2)_{\min}$. The phase retardation, gray scale coded from 0 to π , is given by the rotation angle of the matrix $\mathbf{R}$. Currently, a single rotation matrix is calculated at each depth in the sample, which assumes that the optical axis is constant. In a more advanced approach, the rotation matrix could be calculated between consecutive Stokes vectors along a depth profile, which would take into account the variation of the orientation of the optic axis with depth. The total encountered birefringence would be the sum of the absolute values of the consecutive rotation angles. However, the presence of speckle noise would likely lead to a significant over estimation of total phase retardation.

5. MEASUREMENTS

To demonstrate the technique of polarization sensitive PSOCT, a 2 mm section of human forearm skin was imaged. The RSOD was driven by a 400 Hz triangular waveform, while the polarization modulator was driven by a 200 Hz four-step waveform yielding a scan rate of 800 lines/s and a depth range of 2 mm in tissue ($n = 1.4$). The sample probe was mounted on a linear translation stage that moved with a velocity of 1 mm/s. Total scan time was 2 seconds. Suppressing the surface reflection of the tissue by pressing the forearm against a glass slide with index matching fluid in between allowed efficient utilization of the 65 dB dynamic range of the A/D board. Modulating the incident light over four polarization states produced 16 images (I , Q , U and V for each input state). These were averaged in pairs as discussed in the section on data analysis. The first image, Fig. 5(a), is an average of the reflectivity images of the four different input states. The second image, Fig. 5(b), is a phase retardation map of the tissue calculated from the two vectors S_1 and S_2 . The phase retardation image reveals birefringent structures at a depth of 300 μm below the surface, which we attribute to the presence of collagen in the skin. Fig. 5(c), shows the inner product $S_1 \cdot S_2$ of the two Stokes vectors, gray scale coded between -1 and $+1$. In the absence of dichroism in the sample, the Stokes vectors of the reflected light for the two averaged input states S_1 and S_2 are expected to remain orthogonal, resulting in a uniformly gray image. The extent to which structures are visible in the image is a measure of the dichroism present in the sample. As can be seen from Fig. 5(c), dichroism is not a dominant effect in human skin. The remaining six images Fig. 5(d – i) depict the Stokes parameters Q , U and V of the two averaged polarization states S_1 and S_2 .

SUMMARY

In conclusion, we have demonstrated a fiber-based PSOCT system capable of measuring birefringence and dichroism in *in vivo* human skin. Complications arising from polarization mode dispersion and varying stress-induced birefringence in the single-mode optical fiber were overcome by rapid, interlaced measurements of four incident polarizations which yielded two sets of averaged Stokes vector images. The phase retardation at each depth was calculated from these images assuming a single axis of birefringence. Possible biomedical applications include burn depth determination and early glaucoma detection through retinal nerve fiber layer thickness measurements.

ACKNOWLEDGMENTS

The authors would like to gratefully acknowledge the Whitaker Foundation, Office of Naval Research, Institute of Arthritis, Musculoskeletal and Skin Diseases and the National Center for Research Resources at the National Institute of Health, the U.S. Department of Energy and the Beckman Laser Institute Endowment for the support of this research.

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