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Fiber-optic catheter-based polarization-sensitive OCT for radio-frequency ablation monitoring

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Abstract

An all-fiber optic catheter-based polarization-sensitive optical coherence tomography system is demonstrated. A novel multiplexing method was used to illuminate the sample, splitting the light from a 58.5kHz Fourier-domain mode-locked laser such that two different polarization states, alternated in time, are generated by two semiconductor optical amplifiers. A 2.3mm forward-view cone-scanning catheter probe was designed, fabricated, and used to acquire sample scattering intensity and phase retardation images. The system was first verified with a quarter-wave plate and then by obtaining intensity and phase retardation images of high-birefringence plastic, human skin *in vivo*, and untreated and thermally ablated porcine myocardium *ex vivo*. The system can potentially *in vivo* image of the cardiac wall to aid radio-frequency ablation therapy for cardiac arrhythmias.

Cardiac arrhythmias afflict millions of patients in the United States, resulting in frequent hospitalizations and high medical costs [1]. Catheter-based radio-frequency ablation (RFA) through percutaneous access is commonly practiced in interventional electrophysiology therapeutic procedures to treat cardiac arrhythmias that are not responsive to anti-arrhythmia drugs [2]. RFA creates a thermal lesion to destroy tissue involved in abnormal electrical conduction in order to restore normal conduction patterns. Currently, monitoring of the RFA lesion formation is accomplished only through indirect measures, such as tissue temperature, impedance, and electrograms [2]. Direct imaging feedback at the catheter tip may improve RFA procedures by confirming catheter contact, identifying tissue structures, and confirming ablation lesion formation. We have previously demonstrated that catheter-based optical coherence tomography (OCT) has the potential to provide such guidance [3-7] and the loss of birefringence in the heart wall is a strong marker of ablation lesion formation which can be detected by conventional single-channel OCT. By incorporating polarimetry techniques, polarization-sensitive optical coherence tomography (PSOCT) can provide phase retardation images based on tissue birefringence in addition to scattering intensity images [8-14]. Therefore, PSOCT should detect RFA lesion formation more accurately and robustly than conventional OCT alone.

PSOCT for RFA monitoring will require a fiber-optic catheter scanner. Although catheter-based PSOCT is challenging due to fiber motion, previous work has demonstrated that it is feasible [15-17]. Furthermore, a PSOCT system for clinical use should be simple, robust, and portable. In order to minimize polarization cross-talk, virtually all previously demonstrated PSOCT systems make use of free-space optics (e.g. polarizing beam splitter (PBS) cubes) [9-11, 16] in their detection units and often in their illumination units [10, 11]. Many also make use of polarization-maintaining (PM) fiber [eg. 10]. The use of free-space components requires coupling light out of and back into fibers, and the use of PM fiber requires careful alignment of the fiber orientation. However, the use of these techniques makes alignment more difficult and makes the instrument more sensitive to vibration, which could be an obstacle for clinic application. In this letter, we describe a catheter-based PSOCT system using all fiber-optic components and conventional single mode fiber, including a novel method of modulating the polarization state of the light source.

Fig. 1 shows the schematic of the fiber-optic catheter-based PSOCT system. The light source is a Fourier-domain mode-locked (FDML) laser [18]. The FDML laser has a center wavelength of 1310nm, a bandwidth of 100nm, and a sweep frequency of 58.5kHz. The cavity semiconductor optical amplifier (SOA) is square-wave modulated and only the backward scan of the frequency-tunable filter is used, so the output has a duty cycle of about 50%. The FDML laser cannot be directly used for PSOCT without first being polarized due to chromatic polarization effects [19]. Here, we make use of polarization-sensitive SOAs operating in saturation to force the output light from the FDML laser into a linearly polarized state. The light from the FDML is divided by a 50/50 coupler and sent to a pair of booster polarization-sensitive SOAs. One SOA is delayed by 8.5 microseconds by a coil of fiber such that the two booster SOAs produce alternating laser sweeps. Polarization controllers (PCs) are placed before the booster SOAs to optimize the laser amplification, while an additional PC is placed after each of the booster SOAs to manipulate the alternating laser sweeps into two different linear polarization states at 0 degrees and 45 degrees (90 degrees from each other on the Poincare sphere)[8]. Afterwards, a 50/50 fiber coupler is used to recombine the light from the 2 booster SOAs. The average power output from the coupler is 16mW. This passive multiplexing of alternating polarization states using a buffer fiber spool does not require synchronization of any active component with A-scan data acquisition. This method provides a simple approach to implement all-fiber optic PSOCT with an FDML laser.

The light is then split by a 90/10 coupler. The 10% is used for a calibration interferometer to provide the signal used to linearize the spectral interferogram, while the 90% is connected to an unbalanced Mazh-Zehnder OCT interferometer using circulators in the reference and sample arms as previously described.[20]

The detection unit of the PSOCT system consists of 2 polarization controllers, 2 fiber-based polarizing beam splitters (POBS-13-C-1-7-2, AFW Technologies), and 2 balanced detectors (1817, New Focus). The laser and OCT engine are synchronized and controlled by a computer. Stokes vectors and Mueller calculus are used to calculate the phase retardation map of the birefringent samples. The PSOCT system is configured to acquire OCT images at 50 frames per second with 1000 A-lines per frame.

For calibration, SOA1 was turned off so that all the light was from SOA2 and PC2 was adjusted to maximize output from SOA2. A linear polarizer at 90° and a power meter were placed at the sample position, and the measured power was minimized by adjusting PC4 to ensure that 0° linearly polarized light was output from the SOA2 channel. Similarly, maximum output of 45° linearly polarized light from the SOA1 channel was achieved by setting the linear polarizer to 135°, and adjusting PC1 and PC3. This ensured that the OCT interferometer was illuminated by alternating laser sweeps of 2 states of polarization (SOPs) with a known relationship to each other. PC5 was used to balance the reference light intensity between the two sample polarization states. PC6 and PC7 are used eliminate the return fiber birefringence affect from the probe to the detector to the system. SOA1 was turned off first so only 0 degree linearly polarized light was generated, and a silver mirror was imaged. The output of balanced detector1 was maximized by adjusting PC6 to align the optical axis of PS splitter1. The optical axis of PS splitter2 was aligned in the same way by adjusting PC7.

The Mueller matrix of a general birefringent sample with a slow axis oriented at θ and phase retardation ϕ is:

$$M(\theta, \phi) = \begin{bmatrix} 1 & 0 & 0 & 0 \\ 0 & \cos^2 2\theta + \cos\phi \sin^2 2\theta & (1 - \cos\phi) \sin 2\theta \cos 2\theta & -\sin\phi \sin 2\theta \\ 0 & (1 - \cos\phi) \sin 2\theta \cos 2\theta & \sin^2 2\theta + \cos\phi \cos^2 2\theta & \sin\phi \cos 2\theta \\ 0 & \sin\phi \sin 2\theta & -\sin\phi \cos 2\theta & \cos\phi \end{bmatrix}. \quad (1)$$

The normalized Stokes vectors of 0° and 45° linearly polarized light are:

$$S^0 = \begin{bmatrix} 1 & -1 & 0 & 0 \end{bmatrix}^T, \text{ and} \quad (2)$$

$$S^{45} = \begin{bmatrix} 1 & 0 & 1 & 0 \end{bmatrix}^T. \quad (3)$$

The Stokes vectors of the output light resulting from 0° and 45° linearly polarized light interacting with the birefringent sample are:

$$S_{out}^0 = M((\theta, \phi)) S^0, S_{out}^{45} = M((\theta, \phi)) S^{45}. \quad (4)$$

Considering Eqs. (1), (2), and (3), it can be shown that :

$$S_{1out}^0 = -(\cos^2 2\theta + \cos\phi \sin^2 2\theta), \text{ and} \quad (5)$$

$$S_{2out}^0 = -S_{1out}^{45} = -(1 - \cos\phi) \sin 2\theta \cos 2\theta, \quad (6)$$

where S_{1out}^0, S_{2out}^0 are the second and third elements of S_{out}^0 and S_{1out}^{45} is the second element of S_{out}^{45} . Solving Eqs. (5) and (6) for ϕ in term of Stokes vectors yields:

$$\phi = \arccos \left(\frac{S_{3out}^0}{1 + S_{1out}^0} - 1 \right), \quad (7)$$

Where $S_{3out}^0 = 1 - S_{1out}^0 - S_{2out}^0$. Since the light passes through the sample twice (in and out), the net sample phase retardation is $\phi/2$

We will refer to the A-scans recorded by the 2 balanced detectors (H and V) for 0° and 45° linearly polarized illumination light, respectively, as H_0, V_0, H_{45} , and V_{45} . Therefore, S_{1out}^0 can be calculated at each depth in the A-line from the definition of Stokes vectors as

$S_{1out}^0 = H_0^2 - V_0^2$. Also, S_{2out}^0 can be calculated as $S_{2out}^0 = -S_{1out}^{45} = -(H_{45}^2 - V_{45}^2)$. Thus, it is useful to choose 0 degree and 45 degree linearly polarized light for illumination. The phase retardation is estimated as a function of depth for each A-scan in the image using Equation (7).

A bench-top scanner with a pair of galvanometric mirrors and a lens was first used to test the PSOCT system. Fig. 2 illustrates the combined intensity and phase retardation images of the skin at the proximal interphalangeal joint of the human little finger, averaged over 5 frames. The intensity image in Fig. 2 (a) shows the well-characterized layers that are characteristic of OCT images of human thick skin: stratum corneum, epidermis, and dermis. The phase retardation image in Fig. 2 (b) represents the net phase retardation using a color scale coding 90° in red and 0° in blue, and shows patterns of varying retardation within the epidermis and dermis, similar to what has been shown previously in human thick skin using other PSOCT methods [10].

An OCT catheter based on our previously described cone-scanning forward-viewing concept [7] was fabricated for the PSOCT system. Fig. 3 illustrates the design and a photograph of the probe. The light beam conducted through the optical fiber is first focused by a gradient-index lens and then deflected by a Risley prism. The light spot at focus was measured to be about $16 \mu\text{m}$ at FWHM. All rotating parts were housed in a biocompatible plastic sheath and the light passes through a glass window at the end of the sheath. A miniature fiber-optic rotary joint (FORJ RFCX-131-28, Princetel, Inc.) was used to couple the light from the sample arm of the OCT interferometer to the probe. The sapphire tube controls the position of the glass window and serves as a bearing to facilitate rotation. The sheath is disposable and can be easily replaced. The finished OCT catheter has an outer diameter of 2.3 mm (7.5 F) and a length of 2 m .

The catheter-based PSOCT system, operating in real time, was preliminarily verified by imaging a 1310nm quarter-wave plate (WPQ05M-1310, Thorlabs Inc). The phase retardation of the quarter-wave plate was measured by the catheter PSOCT system while rotating through a full rotation. Compared to the specification of 90° , the averaged phase retardation of the resulting 1000 A-scans is 77.7° with standard deviation of 5.4° . The

under-estimation is expected for at least three reasons. First, the sample was probed by a wide-spectrum swept laser with FWHM of 100nm and centered at 1310nm, not by a single-wavelength laser. Without consideration of the dispersion of the wave plate, the phase retardation difference between 1260nm and 1360nm light is 7.6%. This difference will affect the signal detected and then affect the calculated phase retardation. Second, the result is an average from a B scan (1000 A-scans). Due to catheter rotation, noise is introduced and the illumination polarization states may deviate from 0 and 45 linear states. Because only values of phase retardation between 0° and 90° are produced by the function in Eq. (7), any noise and errors can only result in under-estimation of the true phase retardance [15]. Third, the light is incident on the sample at 9.5 degrees because of the cone-scanning, this will also contribute to the bias [14]. However, the task of detecting loss of tissue birefringence does not rely on quantifying phase retardance, so we consider this bias to be acceptable.

Next, the catheter PSOCT system was used to image a high-birefringence plastic sample and human finger skin. Figs. 4(a) and 4(b) illustrate the combined intensity and phase retardation images of the plastic sample. The phase retardation of the plastic sample cycles periodically from red to blue (0° to 90°), and the birefringence cycles can be observed clearly. Figs. 4(c) and 4(d) illustrate the combined intensity and phase retardation images of the human finger. The images are comparable to the images recorded using the bench-top scanner (Fig. 2). The phase retardation images using the rotating catheter are noisier than those from the bench-top scanner, as expected, but the birefringence signal is still clearly observed.

Finally, we conducted an *ex vivo* experiment demonstrating the feasibility of monitoring RF ablation of myocardial tissue with the system. Left atrium cardiac wedges from a freshly excised swine heart were used for the experiment. And a clinical RFA generator (Maestro 3000, Boston Scientific) with a 7Fr, 4mm tip catheter (Blazer II, Boston Scientific) was used to create lesions by applying RF energy for 20 seconds on the tissue.

Fig. 5 shows fiber-based PSOCT images of left atrium wedges from swine heart. Fig. 5 (a), (c), (e) are the combined intensity, single channel, and phase retardance images before RFA. Fig. 5 (b), (d), (f) are the combined intensity, single channel intensity and phase retardance images after RFA, taken at the edge of the RFA lesion such that the sides show untreated tissue and the middle shows the lesion. As expected, the untreated myocardium exhibits significant phase retardance, while the birefringence is clearly abolished in the RF-treated tissue. These results are consistent with our previous experiments[3,7] The single channel images are equivalent to conventional OCT and the birefringence before ablation can be observed as intensity fading. Compared with the single channel images, the combined intensity images show little polarization artifact, and the birefringence is more easily observed in the phase retardation images. These results demonstrate the potential of the system for RFA monitoring.

In conclusion, a robust, simple, and portable fiber-optic catheter-based PSOCT system was developed and verified to acquire intensity and phase retardation images of biological tissues, including myocardium. The target application is to aid RFA therapy of cardiac arrhythmia by providing direct feedback at the catheter tip. The intended functions of the instrument presented here are to provide interpretable OCT images and to reliably detect the

loss of birefringence in ablated tissue. Thus absolute accuracy of phase retardance was not the design priority. Instead, the system was designed to optimize portability and ease of use by using only fiber-optic components (which do not have optimal polarization performance). The instrument is appropriate and ready for use in pre-clinical trials. Next steps include improving the phase retardation calculation method [8], and testing the system in living hearts in a large animal study to validate that PSOCT more robustly detects loss of birefringence under RFA exposure in the presence of the motion of the heart beat compared to conventional, single-channel OCT.

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Fig. 1.

Schematic of the catheter PS-OCT system. The system consists of a 58.5kHz FDML laser, 2 SOAs to generate alternating laser sweeps with different polarization states, an unbalanced Mach-Zehnder OCT interferometer, a polarization-diversity detector unit, and a forward-view cone-scanning catheter. (FDML: Fourier-domain mode-locked; SOA: semiconductor optical amplifier; PC: polarization controller; FC: fiber coupler; CIR: optical circulator; PS: fiber based polarizing beam splitter; RJ: rotary joint)

Fig. 2. PSOCT images of the skin at the proximal interphalangeal joint of the human little finger using bench-top linear scanner. (a) combined intensity image. (b) Phase retardation image, with blue and red representing phase retardations of 0 and 90° respectively. The images are averages of 5 frames.

Fig. 3. PSOCT forward-viewing probe. (a) Schematic diagram of the probe. (b) Beam profile of the probe at the focus point. (c) Image of the probe without the plastic sheath, glass window, and sapphire tube compared to a penny.

Fig. 4. Single frame catheter PSOCT images. (a), (b) are combined intensity and phase retardation images of a high birefringence plastic. (c), (d) are single channel intensity and phase retardation images of a human finger.

Fig. 5. Single frame catheter PSOCT images of left atrium wedges from swine heart. (a), (c), (e) are intensity (two-channel), single channel, and phase retardance images before RFA. (b), (d), (f) are intensity, single channel, and phase retardance images after RFA, taken at the edge of the RFA lesion such that the two sides are untreated and the middle is the treated area