Swept source cross-polarized optical coherence tomography for any input polarized light

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Abstract
Cross polarized optical coherence tomography (OCT) offers enhanced contrast in certain pathological conditions. Traditional cross-polarized OCT systems require a defined input polarization and thus require several polarization controlling elements increasing the overall complexity of the system. Our proposed system requires a single quarter wave plate as a polarization controller thus simplifying the system significantly. The majority of cross-polarized OCT systems are spectrometer based, which suffers from slow speed and low signal to noise ratio. In this work, we present a swept source based cross-polarized OCT system that works for any input polarization state. The system was tested against known birefringent materials such as quarter wave plate. Furthermore, biological samples such as finger, nail and chicken breast were imaged to demonstrate the potential of our technique.

Supplementary material for this article is available online

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1. Introduction

Optical coherence tomography (OCT) is a relatively new imaging technique which can image tissue with submicron to few micron axial resolution [1–3] in its native state at the video rate. Since its first demonstration in 1991 [4], OCT has been used in several fields, such as ophthalmology [5], gastroenterology [6], cardiology [7] and non-destructive material inspection [8]. The majority of biological samples presents layered structures and any alteration in this architecture, which could be indicative of a pathological condition, can be easily detected using OCT. Plaque formation in coronary artery [7], development of Barrett’s esophagus in the gastroenterology tract [9] and retinal changes in the eye [5] are a few such examples. OCT by itself provides intensity images of the tissue. Several pathologies that are associated with tissue reorganization do not provide enough intensity contrast for early diagnosis. For such cases, another class of OCT including polarization sensitive OCT (PS-OCT) [10–14] and cross polarized OCT (CP-OCT) has been developed, which is based on the detection of the change in input light polarization state. CP-OCT, which is the focus of this work, has been used for esophageal cancer detection [15], brain tumor imaging [16], bladder cancer detection [17], endomyocardial imaging [18] and dental biofilms imaging [19].

Image analysis in CP-OCT requires two images of the tissue, one with the same polarization as the input light and another in orthogonal polarization state. A majority of CP-OCT systems use spectrometer based hardware and the two images are acquired either sequentially [15] or simultaneously using depth encoding [18]. Simultaneous image acquisition is preferred in CP-OCT as it increases the imaging speed and minimizes the tissue motion artifacts. Unfortunately, because
of their limited spectral resolution, spectrometer-based systems suffer from sensitivity degradation as the signal moves away from zero optical path difference. Therefore, the two images acquired with such systems need to be corrected for the sensitivity degradation. Another class of OCT systems, based on swept source light source, offers better sensitivity and relatively lower degradation along depth [20] and thus are more suitable for CP-OCT imaging. A swept source based CP-OCT system has been commercially developed and extensively used in dental studies [21]. However, this system only provides the cross-polarized and not the co-polarized images of the dental structures, limiting its potential. Nevertheless, whether a swept source based or spectrometer based CP-OCT system, both require some known polarization state of the light over the sample. In previously reported CP-OCT systems, polarization of the light from the light source is controlled by a polarization controlling optics. The well controlled polarized light is then passed through a polarizer to have the light with a known polarization state on the sample. This light is further split into reference and sample signal. The polarization of light in the reference arm is further controlled using a quarter wave plate to obtain cross-polarized images. The requirement of polarization controllers and polarizers makes such systems complex. Furthermore, use of polarizer in previously reported systems does not allow us to adopt a balanced detection scheme, which is known to improve the sensitivity in OCT systems [22].

In this work, we present a swept source-based CP-OCT system employing balanced detection, using only a single quarter wave plate as polarization controlling optical element for any input polarized light. Our system offers background reduced, high signal to noise ratio (SNR), functional imaging of biological samples in a simpler manner.

2. Material and methods

A simple technique is utilized in this work to get co-polarized and cross-polarized images simultaneously for any input polarization state. In a lab reference frame, the Jones matrix for an input light with defined polarization state given $E_{0x}$, $E_{0y}$, when double passing through a quarter wave plate resulting in an output light defined by $E_{x}$, $E_{y}$, can be represented as [23]

$$
\begin{bmatrix}
E_{1x}e^{j\phi_{x}} \\
E_{1y}e^{j\phi_{y}}
\end{bmatrix} = \begin{bmatrix}
\cos \theta & \sin \theta \\
-\sin \theta & \cos \theta
\end{bmatrix}
\begin{bmatrix}
E_{0x}e^{j\phi_{0x}} \\
E_{0y}e^{j\phi_{0y}}
\end{bmatrix},
$$

(1)

where $\theta$ is the angle of quarter wave plate fast axis with respect to the lab polarization frame. For any given input polarization state, one can always find a value of $\theta$, such that the inner product of the Jones matrix of the input light with the output light results in zero. This is the orthogonality condition for two polarized light waves and is given by the following equation [24]

$$
E_0^\dagger E_1 = 0,
$$

(2)

where $E_0^\dagger$ is the complex transpose of the Jones vector of the input electric field and $E_1$ is the Jones vector of the output electric field. This formulation is applied in this work and is shown in the schematic of the cross-polarized OCT system in figure 1. This cross-polarized OCT system is based on a commercially available swept source OCT engine (Axius Technologies, North Billerica, MA, USA). The light from the swept source laser has a sweeping rate of 100 kHz, centered at a wavelength of 1310 nm with a 140 nm scan range and 24 mW output power.

The laser light was coupled to a broadband circulator (Thorlabs, CIR-1310-50-APC, Newton, NJ, USA). Afterward, it was coupled to a collimator and split in free space between reference and sample arm using a non-polarizing (50/50) beam splitter (Thorlabs, BS012, Newton, NJ, USA) NPBS1. The reference light was further split into two parts using another non-polarizing (50/50) beam splitter (Thorlabs, BS012, Newton, NJ, USA) NPBS2. One part of the reference light was reflected back using mirror M1 and had identical polarization as the sample signal. An achromatic quarter-wave plate (Thorlabs, AQWP10M-1600, Newton, NJ, USA) was placed in the other reference arm, before mirror M2, and was rotated to obtain orthogonally polarized light relative to input light wave. Experimentally, the required rotation angle for quarter wave plate to obtain orthogonally polarized light in two reference arms can be found very easily by monitoring the intensity of the interference signal between the two reference arms, as the quarter wave plate is rotated. When the interference intensity is set as minimum, the two reference arms signals are set orthogonal to each other. The light in

![Figure 1. The experimental set-up for the cross-polarized optical coherence tomography system. Circulator (C), non-polarizing (50/50) beam splitter (NPBS), mirror (M), quarter wave plate (QWP), lens (L), detector (D).](image-url)
sample arm (9 mW) was focused using a 25 mm focal length lens L and was scanned in 2D using a Galvo-scanner (Thorlabs, GVS012, Newton, NJ, USA). Reflected reference and sample light were combined back at NBPS1. The interfered signals from NBPS1 were split into two parts. One part followed the input light and was coupled back to the circulator’s third arm connected to detector D1. The other part was connected to detector D2 via collimator and fiber for balanced detection. Detector D1 and D2 formed a balanced detector (3 dB bandwidth is 267.96 MHz, ) enclosed within the OCT engine. The balanced detection scheme is widely used in OCT systems to minimize the effect of random intensity noise from the laser source. Since the interference signals reaching detectors D1 and D2 are 180° phase-shifted from each other, it allows one to subtract the noninterfering signals while adding together the interfering signals coming from the sample and the reference surface. Detailed theoretical analysis for a balanced detection scheme can be found in the work done by Podoleanu [22]. The balanced signal from detectors D1 and D2 were digitized using an onboard 12 bit, 500 MS/s digitizer. The digitized signal was then processed with onboard field-programmable gate array (FPGA) board which included Hamming windowing of the spectrum, background subtraction, Fourier transformation and conversion to JPEG images. These images were transferred to a workstation via an ethernet cable using a custom designed Labview (National Instruments) software. The Labview software only acted as an interface between the OCT engine and host workstation and was responsible for the acquisition of the JPEG images from the FPGA module, displaying the images on the screen and then saving them to the hard drive on the host workstation.

In our system, we used two reference signals which were delayed with respect to each other. This generates depth encoded, co- and cross-polarized images (B-Scans) of the sample when the sample beam is scanned over the sample. Reflectivity $R(z)$ and retardance $\delta$ of the sample can be calculated from the amplitude of co- ($A_{co}$) and cross-($A_{cross}$) polarized images using the following formulas [25, 26]

$$R(z) = \sqrt{A_{co}^2 + A_{cross}^2}, \quad (3)$$

$$\delta = \tan^{-1}\left(\frac{A_{cross}}{A_{co}}\right). \quad (4)$$

The data processing for our system involved the following steps:

1. A single B-Scan containing the depth encoded co- and cross-polarized images were transferred from the FPGA module to the computer memory.
2. Co- and cross-polarized images were separated into individual images.
3. A threshold, twice the value of the standard deviation of the noise floor, was applied on the cross-polarized images such that the background noise could be minimized. This step allows us to generate a clearer retardance image.

4. All zeros in the co-polarized image were replaced with the mean of the noise floor. This helps to avoid undefined values in the retardance calculation.
5. Finally, sample reflectivity and retardance were calculated using equations (3) and (4) respectively.

3. Results

The developed system has an axial resolution of approximately 6.8 μm, the lateral resolution of 30 μm, the sensitivity of 106 dB (1 dB drop-off at 5 mm) and the imaging range of 1.5 mm for co-polarized and cross-polarized channel each. It should be noted that although a full imaging range of the system was 5 mm but only 3 mm was used in order to avoid some fixed noise lines arising from multiple optical components used in the system. The imaging speed of the system is 100,000 axial lines (A-lines) per second and is limited by the scan speed of the laser sweeping rate.

To test the performance of the cross-polarized OCT system, we measured the retardance from a known birefringent material, i.e. a quarter wave plate. For this, a mirror was placed at the sample position and a quarter wave plate was inserted in the sample path. The quarter wave plate was rotated between 0° and 90°. Retardance is calculated using equation (4) and is plotted in figure 2.

We imaged several samples using this CP-OCT system. The images were acquired and saved to a hard drive at a 40 Hz frame rate (2500 A-lines/frame). These images are displayed in figures 3 and 4, where each image contains four different images: the co-polarized, the cross-polarized, the standard intensity and the retardance image.

Since balanced detection was employed in this system, we imaged the finger skin with and without balanced detection as shown in figure 3. As can be seen in figure 3, the balanced detection (lower part) displays an increase in SNR.
of 3 dB compared to the unbalanced detection (upper part). The improvement in the SNR was calculated by measuring the average intensity of the images in the balanced detection with respect to the average intensity of the images in the unbalanced case and then dividing these values by the standard deviation of the noise floor measured just above the images where there was no sample signal. Also, the fixed background noise which appears as straight lines, arising from interference between optical components used in the system, is minimized in case of balanced detection whereas such lines are present in the unbalanced case in figure 3. The use of the 2D galvo scanner allowed us to acquire volumetric data. Movies representing the intensity and the retardance for the volumetric scan of the finger and the nail are provided as supplementary data (Movie 1, Movie 2, Movie 3 and Movie 4), available online at stacks.iop.org/JOPT/22/045301/mmedia.

In order to further test the potential of our system, we imaged chicken breast and nail bed tissue using a balanced detection scheme. Both of these tissues are known to be highly birefringent because of the fibrous structures in the tissue. This can be clearly seen by the intensity bands that we can observe in the (a) co- and (b) cross-polarized images of figure 4. Within the co-polarized images, the bands are observed due to complete rotation of the polarization of the sample signal from co- to cross-polarization and therefore a dark band is observed for certain depth within the sample. For a dark band in the co-polarized image, the signal intensity is transferred into a cross-polarized signal and thus a dark band in the co-polarized image appears as a bright band in the cross-polarized image. The frequency and contrast between alternating intensity bands in the retardance image represent the accumulated birefringence of the material of the tissue such as collagen.
Such images are helpful in determining the birefringent properties of the tissue which can be further used to diagnose certain types of pathological conditions. For instance, retardance images have been used to assess the state of human skin [27], coronary artery [28] and cartilage [29]. The ability of our technique to measure the tissue retardance will be helpful in such studies.

4. Conclusion

In conclusion, we have developed a cross-polarized OCT system that works for any input light polarization state. As compared to the previously reported methods, where several polarization controlling optical components are required, our technique requires only one quarter wave plate as a polarization controlling element, thus simplifying the system significantly. We demonstrate the application of this method by measuring the retardance in known optical components such as a quarter wave plate and in biological samples such as finger, nail and chicken breast. We have also employed the balanced detection scheme, which enhanced the SNR of our measurement by 3 dB. A limitation of our system is that it has to be reset for imaging if the input polarization is changed. Although this limitation is shared by most of polarization and cross-polarization based OCT imaging systems, where the input polarization state is controlled by a polarization controller in order to get optimal polarization state. In these imaging systems, once the optimal polarization state is achieved, the fibers can be fixed without the need for further realignment. If the fibers move afterward for any reason, then the polarization state has to be realigned again using a polarization controller. Setting up our system for imaging is achieved fairly easily by rotating the quarter wave plate and minimizing the interference between two of the reference arm signals. This signal is also tracked every 25 ms by our system and any change in input polarization is easily detected through changes in the intensity of the reference arms signal interference. Further realignment of our system can also be minimized by fixing the fibers similar to polarization sensitive or cross polarized OCT systems.

We believe that this simplified cross-polarized OCT system can be a useful diagnostic tool in pathological studies. For instance, the back scattering signal similar to figure 3 can be used to identify pathological conditions in the esophagus and differentiate between healthy mucosa, neoplastic and scar tissues. Fibrous tissue such as collagen, which is a major constituent of several organs, is highly birefringent. Under pathological conditions such as scarring, cancer and burns, collagen quantity or its arrangement is changed which alters the CP-OCT signal and can be identified with the proposed system. There are other areas such as the brain [16] and the heart [18] where CP-OCT has been applied and our simplified system will help the community in such fields.

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