Fiber-based polarization-sensitive OCT for birefringence imaging of the anterior eye segment

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Abstract: We demonstrate a prototype system of polarization-sensitive optical coherence tomography (PS-OCT) designed for clinical studies of the anterior eye segment imaging. The system can measure Jones matrices of the sample with depth-multiplexing of two orthogonal incident polarizations and polarization-sensitive detection. An optical clock is generated using a quadrature modulator and a logical circuit to double the clock frequency. Systematic artifacts in measured Jones matrices are theoretically analyzed and numerically compensated using signals at the surface of the sample. Local retardation images of filtering blebs after trabeculectomy show improved visualization of subconjunctival tissue, sclera, and scar tissue of the bleb wall in the anterior eye segment.

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References and links

1. D. Huang, E. A. Swanson, C. P. Lin, J. S. Schuman, W. G. Stinson, W. Chang, M. R. Hee, T. Flotte, K. Gregory, C. A. Pulsafito, and J. Fujimoto, “Optical coherence tomography,” Science 254(5035), 1178–1181 (1991).
2. J. A. Izatt, M. R. Hee, E. A. Swanson, C. P. Lin, D. Huang, J. S. Schuman, C. A. Pulsafito, and J. G. Fujimoto, “Micrometer-scale resolution imaging of the anterior eye in vivo with optical coherence tomography,” Arch. Ophthalmol. 112(12), 1584–1589 (1994).
3. M. Wojtkowski, A. Kowalczyk, R. Leitgeb, and A. F. Fercher, “Full range complex spectral optical coherence tomography technique in eye imaging,” Opt. Lett. 27(16), 1415–1417 (2002).
4. Y. Yasuno, V. D. Madjarova, S. Makita, M. Akiba, A. Morosawa, C. Chong, T. Sakai, K.–P. Chan, M. Itoh, and T. Yatagai, “Three-dimensional and high-speed swept-source optical coherence tomography for in vivo investigation of human anterior eye segments,” Opt. Express 13(26), 10652–10664 (2005).
5. Y. Yasuno, M. Yamanari, H. Mori, K. Kawana, Y. Watanabe, M. Miura, A. Miyazawa, T. Oshika, and T. Yatagai, “Clinical examinations of anterior eye segments by three-dimensional swept-source optical coherence tomography,” Proc. SPIE 6426(1), 64260U (2007).
6. C. Kerbage, H. Lim, W. Sun, M. Mujat, and J. F. de Boer, “Large depth–high resolution full 3D imaging of the anterior segments of the eye using high speed optical frequency domain imaging,” Opt. Express 15(12), 7117–7125 (2007).
7. M. V. Sarunic, S. Asrani, and J. A. Izatt, “Imaging the Ocular Anterior Segment With Real-Time, Full-Range Fourier-Domain Optical Coherence Tomography,” Arch. Ophthalmol. 126(4), 537–542 (2008).
8. J. F. de Boer, B. Cense, B. H. Park, M. C. Pierce, G. J. Tearney, and B. E. Bouna, “Improved signal-to-noise ratio in spectral-domain compared with time-domain optical coherence tomography,” Opt. Lett. 28(21), 2067–2069 (2003).
9. R. Leitgeb, C. Hitzenberger, and A. Fercher, “Performance of fourier domain vs. time domain optical coherence tomography,” Opt. Express 11(8), 889–894 (2003).
10. M. Choma, M. Sarunic, C. Yang, and J. Izatt, “Sensitivity advantage of swept source and Fourier domain optical coherence tomography,” Opt. Express 11(18), 2183–2189 (2003).
11. S. Yun, G. Tearney, J. de Boer, N. Ifitima, and B. Bouma, “High-speed optical frequency-domain imaging,” Opt. Express 11(22), 2953–2963 (2003).
12. N. Nassif, B. Cense, B. H. Park, S. H. Yun, T. C. Chen, B. E. Bouma, G. J. Tearney, and J. F. de Boer, “In vivo human retinal imaging by ultrahigh-speed spectral domain optical coherence tomography,” Opt. Lett. 29(5), 480–482 (2004).
13. J. M. Schmitt, S. H. Xiang, and K. M. Yung, “Speckle in Optical Coherence Tomography,” J. Biomed. Opt. 4(1), 95–105 (1999).
14. B. H. Park and J. F. de Boer, “Polarization-Sensitive Optical Coherence Tomography,” in Optical Coherence Tomography: Technology and Applications, W. Drexler and J. G. Fujimoto, eds. (Springer Berlin Heidelberg, 2008), pp. 653–695.
15. M. Pircher, C. K. Hitzenberger, and U. Schmidt-Erfurth, “Polarization sensitive optical coherence tomography in the human eye,” Prog. Retin. Eye Res. 30(6), 431–451 (2011).
16. M. Pircher, E. Götzinger, R. Leitgeb, and C. K. Hitzenberger, “Transversal phase resolved polarization sensitive optical coherence tomography,” Phys. Med. Biol. 49(7), 1257–1263 (2004).
17. M. Yamanari, S. Makita, and Y. Yasuno, “Polarization-sensitive swept-source optical coherence tomography with continuous source polarization modulation,” Opt. Express 16(8), 5892–5906 (2008).
18. Y. Yasuno, M. Yamanari, K. Kawana, M. Miura, S. Fukuda, S. Makita, S. Sakai, and T. Oshika, “Visibility of trabecular meshwork by standard and polarization-sensitive optical coherence tomography,” J. Biomed. Opt. 15(6), 061705 (2010).
19. E. Götzinger, M. Pircher, I. Dejaco-Ruhswurm, S. Kaminski, C. Skorpiik, and C. K. Hitzenberger, “Imaging of Birefringent Properties of Keratoconus Corneas by Polarization-Sensitive Optical Coherence Tomography,” Invest. Ophthalmol. Vis. Sci. 48(8), 3551–3558 (2007).
20. S. Fukuda, M. Yamanari, Y. Lim, S. Hoshi, S. Beheregaray, T. Oshika, and Y. Yasuno, “Keratoconus Diagnosis Using Anterior Segment Polarization-Sensitive Optical Coherence Tomography,” Invest. Ophthalmol. Vis. Sci. 54(22), 1384–1391 (2013).
21. Y. Lim, M. Yamanari, S. Fukuda, Y. Kaji, T. Kiuchi, M. Miura, T. Oshika, and Y. Yasuno, “Birefringence measurement of cornea and anterior segment by office-based polarization-sensitive optical coherence tomography,” Biomed. Opt. Express 2(8), 2392–2402 (2011).
22. S. Fukuda, S. Beheregaray, D. Kasaragod, S. Hoshi, G. Kishino, K. Ishii, Y. Yasuno, and T. Oshika, “Noninvasive Evaluation of Phase Retardation in Blebs After Glaucoma Surgery Using Anterior Segment Optical Coherence Tomography,” Invest. Ophthalmol. Vis. Sci. 55(8), 5200–5206 (2014).
23. M. Yamanari, K. Ishii, S. Fukuda, Y. Lim, L. Duan, S. Makita, M. Miura, T. Oshika, and Y. Yasuno, “Optical Rheology of Porcine Sclera by Birefringence Imaging,” PLoS ONE 7(9), e44026 (2012).
24. S. Nagase, M. Yamanari, R. Tanaka, T. Yasui, M. Miura, T. Iwasaki, H. Goto, and Y. Yasuno, “Anisotropic Alteration of Scleral Birefringence to Uniaxial Mechanical Strain,” PLoS ONE 8(3), e58716 (2013).
25. M. Yamanari, S. Nagase, S. Fukuda, K. Ishii, R. Tanaka, T. Yasui, T. Oshika, M. Miura, and Y. Yasuno, “Scleral optical coherence tomography: measurement by polarization-sensitive optical coherence tomography and ocular biometric parameters of human eyes in vivo,” Biomed. Opt. Express 5(5), 1391–1402 (2014).
26. C. E. Saxer, J. F. de Boer, B. H. Park, Y. Zhao, Z. Chen, and J. S. Nelson, “High-speed fiber based polarization-sensitive optical coherence tomography of in vivo human skin,” Opt. Lett. 25(18), 1355–1357 (2000).
27. S. Jiao, W. Yu, G. Stoica, and L. V. Wang, “Optical-fiber-based Mueller optical coherence tomography,” Opt. Lett. 28(14), 1206–1208 (2003).
28. B. H. Park, M. C. Pierce, B. Cense, and J. F. de Boer, “Jones matrix analysis for a polarization-sensitive optical coherence tomography system using fiber-optic components,” Opt. Lett. 29(21), 2512–2514 (2004).
29. W. Y. Oh, S. H. Yun, B. J. Vakoc, M. Shishkov, A. E. Desjardins, B. H. Park, J. F. de Boer, G. J. Tearney, and B. E. Bouma, “High-speed polarization sensitive optical frequency domain imaging with frequency multiplexing,” Opt. Express 16(2), 1096–1103 (2008).
30. W. Y. Oh, B. J. Vakoc, S. H. Yun, G. J. Tearney, and B. E. Bouma, “Single-detector polarization-sensitive optical frequency domain imaging using high-speed intra-A-line polarization modulation,” Opt. Lett. 33(12), 1330–1332 (2008).
31. Y. Lim, Y.-J. Hong, L. Duan, M. Yamanari, and Y. Yasuno, “Passive component based multifunctional Jones matrix swept source optical coherence tomography for Doppler and polarization imaging,” Opt. Lett. 37(11), 1958–1960 (2012).
32. B. Baumann, W. Choi, B. Potsaid, D. Huang, J. S. Duker, and J. G. Fujimoto, “Swept source/Fourier domain polarization sensitive optical coherence tomography with a passive polarization delay unit,” Opt. Express 20(9), 10229–10241 (2012).
33. Z. Wang, H.-C. Lee, O. O. Ahsen, B. Lee, W. Choi, B. Potsaid, J. Liu, V. Jayaraman, A. Cable, M. F. Kraus, K. Liang, J. Horngger, and J. G. Fujimoto, “Depth-encoded all-fiber swept source polarization sensitive OCT,” Biomed. Opt. Express 5(9), 2931–2949 (2014).
34. J. Xu, L. Huo, J. Li, and X. Li, “Generic real-time uniform K-space sampling method for high-speed swept-source optical coherence tomography,” Opt. Express 18(9), 9511–9517 (2010).
35. E. Z. Zhang and B. J. Vakoc, “Polarimetry noise in fiber-based optical coherence tomography instrumentation,” Opt. Express 19(18), 16830–16842 (2011).
36. M. Villiger, E. Z. Zhang, S. Nadkarni, W.-Y. Oh, B. E. Bouma, and B. J. Vakoc, “Artifacts in polarization-sensitive optical coherence tomography caused by polarization mode dispersion,” Opt. Lett. 36(8), 923–925 (2013).
1. Introduction

Optical coherence tomography (OCT) is an imaging modality to measure the two or three-dimensional structure of biological samples [1]. In addition to the retinal imaging that is the most popular application of OCT, an anterior segment OCT (AS-OCT) has also been developed to observe the morphometric properties of the anterior eye segment. The technique of the interferometer for AS-OCT was translated from time-domain OCT [2] into spectral-domain OCT and swept-source OCT (SS-OCT) or optical frequency domain imaging (OFDI) [3–7] because of advantages in the sensitivity [8–10] and ability of high-speed imaging [11,12].

Conventional OCT utilizes only scattering intensity to form images, and speckle contrast in a broad sense is the sole agent to distinguish different tissues [13]. Polarization-sensitive OCT (PS-OCT) has been developed to extend the function of OCT by exploiting the full nature of polarized light [14,15]. Previous studies of the anterior eye segment using PS-OCT showed promising potential in birefringence imaging for enhancing image contrast in normal tissue [16–18] and detecting abnormal birefringence in keratoconus corneas [19,20] and scar tissue [21,22]. In addition, it has been shown that scleral birefringence had statistically significant correlations with scleral elasticity and intraocular pressure in some studies [23–25], suggesting that scleral ultrastructure is partially related to the rheology of the eye. These reports have shown a potential of PS-OCT regarding the application of the anterior eye segment that cannot be obtained by conventional OCT.

Fiber-based PS-OCT is useful to facilitate optical alignment and to enable a flexible probe design for the measurement of Jones matrices or Stokes parameters [26–28]. In order to measure these parameters without reducing the effective A-scan rate in SS-OCT, frequency- or depth-multiplexing methods have been developed [17,29–32]. Wang et al. [33] applied electric frequency doubling of the sampling clock to extend the axial measurement range of the depth-encoded PS-OCT. Although the electric frequency doubling is effective, its
applicability is limited by the bandwidth of the frequency multiplier. If the frequency-swept laser has high nonlinearity of the optical frequency sweep, the multiplied clock can be distorted. In the case of frequency multiplexing with an electric 90-degree phase shifter [34], the applicability is also limited by frequency dependence of the electric phase shifter.

Another challenge in PS-OCT is wavelength-dependent polarization properties of optical components. Recent studies have suggested that polarization mode dispersion (PMD) increases polarimetric noise in PS-OCT [35,36]. Some methods have been developed to overcome this issue using various approaches [37–39].

In this paper, we report on our depth-multiplexed PS-OCT using a sampling clock generated by a combination of optical and electric components. We also show our methods to correct wavelength-dependence of polarization-sensitive devices in the system using signals at the surface of the sample and to reduce speckle noise of the Jones matrix using a coherent spatial Gaussian filter. Local retardation images of filtering blebs after trabeculectomy are demonstrated as a clinical application of our system.

2. PS-OCT system

2.1 Interferometer

The interferometer of our system employs the depth-multiplexing method with passive components [31], but the details were refined and adapted for imaging of the anterior eye segment. Figure 1(a) shows a schematic of the interferometer. Light source is a frequency-swept laser (HSL-20, Santec, Aichi, Japan), which has a center wavelength of 1297 nm, a wavelength range of 112 nm, a sweep rate of 50 kHz, a duty cycle of 51%, and an average output power of 34.0 mW. All of the optical fiber used in this system is single mode fiber (SMF). 10% of the output light is directed to a trigger/clock generator, which will be described below. 90% of the output light is directed at a Mach-Zehnder interferometer. In the sample arm, a polarization delay unit (PDU) was built to apply different optical delays for two orthogonal polarizations. The beam is divided by a polarizing beam splitter (PBS), reflected by right-angle prisms, combined at the same PBS, and coupled to SMF. Since the single PBS ensured the same tilt angle of the PBS for both the incident path and the return path, the beam direction after combining the two beams from two prisms can be easily aligned. This design reduced the effort of the alignment and the implemented size of the PDU, which was 150 × 200 mm² using ready-made components. After the PDU, the light passes through a circulator (Oplink Communications Inc., CA, USA), and is directed to a scanner illuminating a sample with an averaged optical power of 7.71 mW. The optical power loss between the light source and the probe was mainly due to the free space implementation of the PDU and the connections of the SMF patch cables. The scanned position on the sample is monitored by a CMOS camera through a dichroic mirror using light-emitting diodes at a wavelength of 940 nm. Backscattered light is coupled to SMF, and directed to a polarization-sensitive detection arm. In the reference arm, the light passes through a variable delay line, and is directed to the polarization-sensitive (PS-) detection unit. In the PS-detection unit, two orthogonal polarizations are separated and detected individually. Fiber Bragg grating (FBG, FBG-SMF-1264-80-0.2-A-{2}60F/E,L = 1M, Tatsuta Electric Wire & Cable Co., Ltd, Osaka, Japan) was used in front of one of two ports of each balanced receiver to embed a peak signal at a constant wavelength in the interference signal. These signals are detected by two balanced receivers that have a bandwidth of 200 MHz (C12792-2555(X), Hamamatsu Photonics, Shizuoka, Japan), filtered by a low-pass filter (SLP-300 + , Mini Circuits, NY, USA), and digitized by a high-speed digitizer (ATS9350, Alazartech, QC, Canada). The low-pass filter was selected so that the nonlinear group delay of the filter did not influence the signal phase of OCT, and was used to eliminate aliasing of high-frequency signals that passed the balanced receivers over their bandwidth.
A schematic of the trigger/clock generator is shown in Fig. 1(b) and Fig. 1(c). Interference signals were generated by a Mach-Zehnder interferometer with quadrature modulation as shown in Fig. 1(b). A phase shift of 90 deg was applied between horizontal and vertical polarizations in one arm of the interferometer using a quarter waveplate (QWP). Interference signals of horizontal and vertical polarizations were divided by a PBS and detected individually by two receivers (PDB130C, Thorlabs, NJ, USA). These two signals of 125-235 MHz frequency were bandpass-filtered to suppress unused frequency bands, filtered by comparators to binarize signals at zero-crossing points, and combined with a logical XOR gate to double the signal frequency to 250-470 MHz as shown in Fig. 1(c). The XORed signal was bandpass-filtered, amplified, and supplied to the digitizer as an external clock. This electric processing using XOR gate is similar to Xi et al. except the phase shifting method [34]. One output port of the beamsplitter (BS) that was unused for clock generation in Fig. 1(b) was connected to an FBG. The transmitted light from the FBG was used for monitoring the optical power, and the reflected light from it was used as an optical trigger of the digitizer.
Our optical clock enabled an axial measurement range of 15.28 mm for standard OCT imaging. For depth-multiplexed PS-OCT imaging, the optical delay of the PDU was set to be 6.68 mm, and it was the axial measurement range for PS-OCT imaging.

The sensitivity, axial resolution and signal roll-off of the system were 97.6 dB at 3.41 mm depth, 10.1 µm in tissue and $-0.54$ dB/mm, respectively.

### 2.2 Prototype system for clinical studies

All of the interferometer except the probe of the sample arm was installed in the interferometer assembly box shown in Fig. 2. The scanning probe was installed on a scanning head, and it was connected to the sample arm of the interferometer with an SMF. All of the hardware, including a computer and a monitor, were mounted on a motorized optical table. This PS-OCT system was set up in the clinic of Tohoku University Hospital. Software for both the measurement and analysis was developed using LabVIEW (National Instruments, TX, USA) so that the system can be operated by orthoptists.

![Fig. 2. A photo of the installed PS-OCT system.](image)

### 3. Signal modeling and processing

#### 3.1 Signal pre-processing

Volumetric raw spectra were acquired using the system described in Section 2. Each spectrum has 3200 pixels that were sampled based on the optical clock. DC offset of the spectrum was calculated by 10th-order polynomial fit to each average of B-scan spectra and subtracted from each spectrum.

In principle, the spectrum has no jitter in the sampled timing owing to the optically-generated trigger and sampling clock. In practice, however, we observed the jitter that is in integral multiples of the sampling clock between the horizontal and vertical detection channels and among A-scans. Although the cause of this jitter is unclear, we speculate that it may be because of a stability issue in the electric analog circuit of the analog-to-digital conversion with the external clock in the digitizer. To resolve this issue, a peak signal was superposed in each spectrum at a wavelength of 1264 nm using the FBGs in the PS-detection unit. This approach is an extension of the method by Choi et al. [40] that compensated for the timing jitter among A-scans of a single detection channel. The following processing was applied in our compensation method; 1) 128 pixels were extracted from each spectrum around the peak signal; 2) a number of the pixels were increased by 16 times using zero-padding technique; 3) the center of rising and falling edges of the peak that were detected at half maximum of the peak was defined as the peak position; 4) all spectra in the volumetric data were shifted to align the peak positions in integral multiples of the original 128-pixel resolution.
Subsequently, the spectrum was apodized by Hanning window and transformed by
discrete Fourier transform (DFT). The positive-frequency range was extracted to obtain the
complex OCT signal. The depth-multiplexed signals were demodulated by shifting the axial
depth in a precision of the discrete pixel resolution, which was 9.55 µm/pixel. Fixed pattern
noise in these complex OCT signals was removed by subtracting the complex median at each
depth in each B-scan [41]. These OCT signals were transformed into spectral domain by
inverse DFT for the successive processing shown in the following section.

3.2 Modeling of the electric fields with Jones formalism

In depth-multiplexed PS-OCT, four signals are detected from each A-scan using the PDU and
two detectors of H and V channels. We start from an equation that was previously described
by Braaf et al. in Eq. (8) of [37] to model these electric fields;

$$
\mathbf{E}(k) = \begin{bmatrix}
E_x(k), \\
E_y(k)
\end{bmatrix} = \begin{bmatrix}
E_{H1}(k) & E_{H2}(k) \\
E_{V1}(k) & E_{V2}(k)
\end{bmatrix}
\times \begin{bmatrix}
R_s(k) & 0 \\
0 & R_t(k)
\end{bmatrix}
\times \begin{bmatrix}
J_{out}(k)J_{s}(k)J_{in}(k)\left[\alpha(k)\beta(k\Delta + \gamma)\exp(k\gamma + \rho(k))\right] \\
0
\end{bmatrix}
\mathbf{e}_n(k),
$$

(1)

where $k$ is an optical wavenumber; $E_{H1}(k), E_{V1}(k)$ and $E_{H2}(k), E_{V2}(k)$ are horizontal or
vertical components of depth-multiplexed electric fields $E_x(k)$ and $E_y(k)$, respectively;
$R_s(k)$ and $R_t(k)$ are horizontal and vertical components of the reference electric field at the
detectors; $J_s(k)$ is a Jones matrix of an optical path from the fiber leaving the PDU up to the
sample’s surface; $J_{out}(k)$ is a round-trip Jones matrix of the sample; $J_{in}(k)$ is a Jones matrix
of an optical path from the sample’s surface up to the detector; $\alpha(k)$ is an amplitude
difference between the two electric fields at the output port of the PDU; $\beta(k\Delta)$ is a depth-
dependent signal decay at a depth of $k\Delta$; $\gamma$ is a depth displacement introduced by the PDU
to the horizontal polarization; $\rho(k)$ is the phase difference between the horizontal and vertical
field components of the light source; $\mathbf{e}_n(k) = c_n(k)\exp(ik\Delta)$ is a common electric field component,
where $c_n(k)$ is a common amplitude of the interfered signals. Since $\gamma$ is already demodulated
as described in Section 3.1, it is replaced with a residual phase $\gamma'$ after the demodulation.
Equation (1) is rewritten with a new symbol $\mathbf{E}_{\text{sample}}(k)$ instead of $\mathbf{E}(k)$ as

$$
\mathbf{E}_{\text{sample}}(k) = \begin{bmatrix}
R_s(k) & 0 \\
0 & R_t(k)
\end{bmatrix}
\times \begin{bmatrix}
J_{out}(k)J_{s}(k)J_{in}(k)\left[\alpha(k)\beta(k\Delta + \gamma)\exp(k\gamma + \rho(k))\right] \\
0
\end{bmatrix}
\mathbf{e}_n(k),
$$

(2)

Note that the terms of the depth-dependent signal decay, $\beta(k\Delta)$ and $\beta(k\Delta + \gamma')$, are not
compensated at this step. $J_{out}(k)$ in Eq. (2) can be rearranged as

$$
\mathbf{E}_{\text{sample}}(k) = \begin{bmatrix}
R_s(k)e^{i\phi(k)} & 0 \\
0 & R_t(k)
\end{bmatrix}
\times \begin{bmatrix}
J_{out}(k)J_{s}(k)J_{in}(k)\left[\alpha(k)\beta(k\Delta + \gamma)\exp(k\gamma + \rho(k))\right] \\
0
\end{bmatrix}
\mathbf{e}_n(k),
$$

(3)

where $J_{out}'(k)$ is a Jones matrix of an optical path from the sample’s surface up to the input
fiber tip of the PS-detection unit, and $\phi(k)$ is the phase difference between the horizontal and vertical
components due to the PS-detection unit, which includes a mismatch of the optical
path lengths between the horizontal and vertical components of the electric fields from the
PBS to the detectors.
3.3 Compensation of the signal decay and k-dependence

In this section, we show our compensation method using the sample’s surface. The depth position of the speckle peak at the sample’s surface in a B-scan of the volumetric data is detected using an OCT intensity image. A narrow Gaussian window that has a standard deviation of 3 pixels (28.7 µm in air) and the center at the surface was multiplied to the complex OCT A-scan profiles that were calculated in the pre-processing step of Section 3.1, and transformed into spectral domain by inverse DFT. Assuming \( J_{ik} \) is an identity matrix at the sample’s surface, the electric fields of the surface are derived from Eq. (3) as

\[
E_{\text{surf}}(k) = \begin{bmatrix} E_{H1\text{surf}}(k) & E_{H2\text{surf}}(k) \\ E_{V1\text{surf}}(k) & E_{V2\text{surf}}(k) \end{bmatrix} = \begin{bmatrix} R_{H}(k)e^{i\phi(k)} & 0 \\ 0 & R_{V}(k) \end{bmatrix} \begin{bmatrix} J'_{\text{surf}}(k)J_{ik}(k) \end{bmatrix} \begin{bmatrix} \alpha(k)\beta(\Delta\varepsilon+\gamma)e^{i(k\gamma'+\rho(k))} \\ 0 \end{bmatrix} \begin{bmatrix} \beta(\Delta\varepsilon) \end{bmatrix} e^{i\chi}, \tag{4}
\]

Equation (4) is demodulated by multiplying a complex conjugate of the normalized \( E_{H1}(k) \) and is averaged in the B-scan to increase reliability. The averaged electric fields are described as

\[
E'_{\text{surf}}(k) = \begin{bmatrix} R_{H}(k)e^{i\phi(k)} & 0 \\ 0 & R_{V}(k) \end{bmatrix} \begin{bmatrix} J'_{\text{surf}}(k)J_{ik}(k) \end{bmatrix} \begin{bmatrix} \alpha(k)\beta(\Delta\varepsilon+\gamma)e^{i(k\gamma'+\rho(k))} \\ 0 \end{bmatrix} \begin{bmatrix} \beta(\Delta\varepsilon) \end{bmatrix} e^{i\chi}, \tag{5}
\]

where \( e^{i\chi} = e^{i\Delta\varepsilon}E_{H1\text{surf}} / |E_{H1\text{surf}}| \) calculated from Eq. (4), and the superscript * means the complex conjugate. Since \( e^{i\Delta\varepsilon} \) of \( e_{ik}(k) \) in Eq. (4) is cancelled, \( e^{i\chi} \) is a constant common phase and it does not influence the compensation process in the following. Matrix elements of Eq. (5) are individually described for later use as

\[
E'_{\text{surf}}(k) = \begin{bmatrix} E'_{H1\text{surf}}(k) & E'_{H2\text{surf}}(k) \\ E'_{V1\text{surf}}(k) & E'_{V2\text{surf}}(k) \end{bmatrix}. \tag{6}
\]

Assuming that \( J_{ik}(k) \) and \( J'_{\text{surf}}(k) \) are unitary matrices and their k-dependence is negligibly small in our system and using \( J'_{\text{surf}}J_{ik} := [J_{00}, J_{01}, J_{10}, J_{11}] \), relative values of k-dependent parameters in Eq. (5) are derived;

\[
|q_{HP}(k)| := \left| \frac{E'_{H2\text{surf}}(k)E''_{V1\text{surf}}(k)}{E'_{H1\text{surf}}(k)E''_{V2\text{surf}}(k)} \right|^{\frac{1}{2}} = \left| \frac{R_{H}(k) - R_{V}(k)}{R_{H}(k)} \right|, \tag{7}
\]

\[
|q_{H2}(k)| := \left| \frac{E'_{V2\text{surf}}(k)E''_{H1\text{surf}}(k)}{E'_{V1\text{surf}}(k)E''_{H2\text{surf}}(k)} \right|^{\frac{1}{2}} = \frac{\beta(\Delta\varepsilon)}{\alpha(k)\beta(\Delta\varepsilon+\gamma)}, \tag{8}
\]

\[
\arg \left\{ q_{HP}(k) \right\} := \arg \left\{ \frac{E'_{H2\text{surf}}(k)E''_{V1\text{surf}}(k)}{E'_{H1\text{surf}}(k)E''_{V2\text{surf}}(k)} - \frac{E'_{V2\text{surf}}(k)E''_{H1\text{surf}}(k)}{E'_{V1\text{surf}}(k)E''_{H2\text{surf}}(k)} \right\} = \arg \left\{ R_{V}(k)R_{H}(k)e^{i\phi(k)} \left( \frac{\alpha(k)\beta(\Delta\varepsilon+\gamma) + \beta(\Delta\varepsilon)^2}{2} \right) \right\} \tag{9}
\]

arg \{ q_{12}(k) \} = \arg \left( E'_{V1surf}(k)E''_{H1surf}(k) - E'_{H2surf}(k)E''_{V2surf}(k) \right) \\
= \arg \left( \beta(\Delta\omega)\alpha(k)\beta(\Delta\omega + \gamma)e^{-i(k'y' + \rho(k))} \left[ |R_H(k)|^2 + |R_V(k)|^2 \right] J_{10}J'_{00} \right) \quad (10)

= -(k'y' + \rho(k)) + \eta,

where \( \zeta = \arg \{ J_{10}',J_{00}' \} \) and \( \eta = \arg \{ J_{00}',J_{00}' \} \). Equations (7)-(10) are used to construct the following diagonal matrices as

\[
Q_{HF}(k) := \begin{bmatrix} 1 & 0 \\ 0 & q_{HF}(k) \end{bmatrix},
\]

\[
Q_{12}(k) := \begin{bmatrix} 1 & 0 \\ 0 & q_{12}(k) \end{bmatrix}.
\]

Equation (3) is rearranged using Eqs. (11) and (12) as

\[
E_{sample}(k) = Q_{HF}(k)J_{out}^*(k)J_{in}'(k)Q_{12}(k)e_{in}'(k),
\]

where

\[
J_{out}^* = \begin{bmatrix} 1 & 0 \\ 0 & e^{-i\zeta} \end{bmatrix} J_{out}^*,
\]

\[
J_{in}' = \begin{bmatrix} 1 & 0 \\ 0 & e^{-i\eta} \end{bmatrix},
\]

\[
e_{in}'(k) = R_H(k)e^{i\eta(k)}\alpha(k)\beta(\Delta\omega + \gamma)e^{i(k'y' + \rho(k))} e_{in}(k).
\]

The \( k \)-dependence of \( E_{sample}(k) \) is corrected using Eqs. (11)-(13) as

\[
\hat{E}_{sample}(k) := Q_{HF}^{-1}(k)E_{sample}(k)Q_{12}^{-1}(k) = J_{out}^*J_{in}'(k)J_{in}'e_{in}'(k).
\]

Depth-resolved Jones matrices that include polarimetric properties of the sample are calculated by applying DFT to Eq. (15) as

\[
\hat{E}_{sample}(z) = J_{out}^*J_{in}(z)J_{in}'e_{in}'(z).
\]

Representative numerical examples of \( E'_{surf}(k) \), \( q_{HF}(k) \) and \( q_{12}(k) \) are shown in Fig. 3. The sample was the human anterior eye segment shown in Fig. 4. Since \( E'_{H1surf}(k) \) was used to demodulate the signals at the sample’s surface, \( \arg \{ E'_{H1surf}(k) \} \) was constant in Fig. 3(b). All other matrix elements of \( E'_{surf}(k) \) showed \( k \)-dependence of the phase that is relative to \( \arg \{ E'_{H1surf}(k) \} \) as shown in Fig. 3(b). The \( \arg \{ E'_{V1surf}(k) \} \) had the spectral phase slope that originated from the time delay between H and V channels and DC offset that depended on fiber bending as characterized by \( J_{out}'J_{in} \). The \( \arg \{ E'_{H2surf}(k) \} \) and \( \arg \{ E'_{V2surf}(k) \} \) had a higher phase slope, which was typical in our system because mechanical perturbation of the PDU in the sample arm was more influential than that of the detection unit. The phase slope of \( \arg \{ E'_{V2surf}(k) \} \) was the highest among the elements of \( E'_{surf}(k) \) in this example, as it had both influences from the PDU and the detection unit. These phase characteristics of \( E'_{surf}(k) \) were outlined by \( \arg \{ q_{HF}(k) \} \) and \( \arg \{ q_{12}(k) \} \) in Fig. 3(d). Similarly, \( k \) and element-dependent amplitudes of \( E'_{surf}(k) \) shown in Fig. 3(a) that were originated from \( k \) and
polarization-dependent optical losses in the PDU and the detection unit were outlined in Fig. 3(c). All of these systematic artifacts were corrected in Eq. (15).

Fig. 3. Numerical examples of $E_\text{surf}^r(k)$, $q_{\text{H1}}(k)$ and $q_{\text{V2}}(k)$. Spectral amplitudes (a) and phases (b) of $E_\text{surf}^r(k)$ elements and spectral amplitudes (c) and phases (d) of $q_{\text{H1}}(k)$ and $q_{\text{V2}}(k)$ are shown. In the graph legends of (a) and (b), H1, V1, H2 and V2 are the abbreviations of matrix elements of Eq. (6). In (a), the amplitudes were normalized by the highest value in the four matrix elements for data visualization.

3.4 Speckle reduction of Jones matrices with coherent spatial Gaussian filter

Speckle has both attributes as a fundamental carrier of information and noise source for OCT [13]. In PS-OCT, Jones matrix averaging methods have been proposed with global phase cancellation to reduce the speckle noise [21,37,42]. In this section, we describe our speckle reduction technique using a coherent spatial Gaussian filter.

The Jones matrix at each pixel in a B-scan has an unpredictable global phase offset because of a coherent combination of densely packed unknown scatterers. Assuming that the sample has a uniform optic axis of birefringence without diattenuation in a small rectangular spatial kernel, the Jones matrix in the kernel can be modeled instead of Eq. (16) as

$$K(n) := e^{\Phi(n)} U_{\text{surf}} \begin{bmatrix} e^{\Delta\phi(n)} \lambda_1 & 0 \\ 0 & e^{\Delta\phi(n)} \lambda_2 \end{bmatrix} U_{\text{in}},$$

(17)

where $\Phi(n)$ is the global phase at $n$-th pixel in the kernel, $U_{\text{in}}$ and $U_{\text{surf}}$ are unitary matrices, $\lambda_1$ and $\lambda_2$ are eigenvalues of $J_s$ at the reference pixel ($n = 0$) in the kernel with a unit absolute value $|\lambda_1| = |\lambda_2| = 1$, $\Delta\phi(n)$ is phase retardation that is relative to the reference pixel.
Parameterizing elements of the general unitary matrices $U_{in}$ and $U_{out}$ in Eq. (17), a relative global phase $\Phi(n) - \Phi(0)$ is derived as

$$\arg \left\{ K_{H1}(n)K_{H1}^*(0) + K_{v1}(n)K_{v1}^*(0) + K_{H2}(n)K_{H2}^*(0) + K_{v2}(n)K_{v2}^*(0) \right\} = \Phi(n) - \Phi(0),$$

where $K(n) = [K_{H1}(n), K_{H2}(n); K_{v1}(n), K_{v2}(n)]$. Note that $\Delta \phi(n), \lambda_1, \lambda_2$ and all elements of $U_{in}$ and $U_{out}$ were eliminated at the right side of Eq. (18). A pixel that has the maximum value of effective signal-to-noise ratio [43] was selected as the reference pixel in the kernel. The global phase was cancelled by multiplying $e^{-i(\Phi(n) - \Phi(0))}$ to Eq. (16).

After the cancellation of the global phase, a weight function of two-dimensional spatial Gaussian was multiplied over the kernel, and the electric fields were spatially averaged. In this averaging process, it was assumed that the sample had uniformly cumulative phase retardation along the depth due to uniform birefringence in the kernel. The vector summations of the cumulative Jones-matrix elements converge to the Jones matrix at the center of the kernel. The kernel moved all over the B-scan pixels, and the same averaging was applied to all pixels. The electric fields after this processing is denoted as $\hat{E}_{\text{sample}}(z)$. Compared to the rectangular weight function that is equivalent to sinc function in spatial frequency space, this Gaussian filtering is effective in reducing speckle noise that has high spatial frequency. The sizes of the rectangular kernel were 5 pixels (34.4 µm in tissue, assuming the refractive index of tissue to be 1.389) in axial direction and 15 pixels (225 µm) in lateral direction. Standard deviations of the spatial Gaussian filter were 1 pixel (6.9 µm in tissue) in axial direction and 3 pixels (45 µm) in lateral direction. It is easy to extend this method to three-dimensional space. In this paper, however, we applied it only in two-dimensional B-scan space, because our scanning protocol was dense only in a lateral direction of the B-scan.

### 3.5 Intensity and local retardation

The intensity of the electric fields was calculated as $\frac{1}{3}(|\hat{E}_{H1}|^2 + |\hat{E}_{v1}|^2 + |\hat{E}_{H2}|^2 + |\hat{E}_{v2}|^2)$, where these parameters are matrix elements of Eq. (16). In order to calculate the local retardation of birefringence, we applied a similar approach to Makita et al. [43] that took axially separated Jones matrices as

$$\hat{E}_{\text{sample}}(z - \frac{\delta z}{2}) \hat{E}_{\text{sample}}(z + \frac{\delta z}{2}) = J_{\text{in}}^{-1}(z) \begin{bmatrix} \lambda_1(\delta z) & 0 \\ 0 & \lambda_2(\delta z) \end{bmatrix} J_{\text{out}}(z) J_{\text{in}}^{-1}(z),$$

where $\delta z$ is a length of the axial pixel separation, $J_{\text{in}}(z)$ is the cumulative optic axis of the sample from the sample’s surface to the depth $z$, $\lambda_1(\delta z)$ and $\lambda_2(\delta z)$ are eigenvalues of the local Jones matrix in the axial range $\delta z$ at the depth $z$. The eigenvalues are calculated as

$$\lambda_1(\delta z) = \frac{T}{2} + \left( \frac{T^2}{4} - D \right)^{\frac{1}{2}},$$

$$\lambda_2(\delta z) = \frac{T}{2} - \left( \frac{T^2}{4} - D \right)^{\frac{1}{2}},$$

where $T$ and $D$ are the trace and determinant of Eq. (19), respectively. The double-pass local retardation per unit depth $r(\delta z)$ is calculated as
The axial pixel separation $\delta z$ was set to be 4 pixels (27.5 $\mu$m in tissue).

4. Results of PS-OCT imaging

One healthy eye and three glaucomatous eyes of four human volunteers were involved in this study. The study protocol adhered to the tenets of the Declaration of Helsinki and was approved by the institutional review board of Tohoku University Graduate School of Medicine. The anterior eye segment was scanned by the PS-OCT prototype system with a raster scan protocol and a radial scan protocol. In the raster scan protocol, a lateral region of $12 \times 12$ mm$^2$ was scanned by 800 \times 128 A-scans. In the radial scan protocol, a lateral range of 12 mm was scanned by 800 A-scans and the B-scan direction was rotated in a counterclockwise direction to cover 360 degrees with 128 B-scans. The best B-scan was selected from these scan patterns to show representative properties of each subject.

4.1 Efficacy of the processing methods in a healthy human eye

The healthy anterior eye segment at the superi or angle of 64-year-old female was measured by PS-OCT, and the processed results are shown in Fig. 4. The OCT intensity image in Fig. 4(b) shows the angle of the anterior chamber without artificial signal fading due to birefringent tissue [18]. The image quality of local retardation images depending on the $k$-dependence correction and the coherent spatial Gaussian filter is shown in Fig. 4(c). A color map of Ametrine was used to display the local retardation images [44]. Without applying both processing methods, polarimetric speckle noise was too high to see any difference between subconjunctival tissue and sclera as shown in Fig. 4(c1). Although the polarimetric speckle noise was reduced by applying the coherent spatial Gaussian filter in Fig. 4(c2), the local retardation image still had noise, which decreased the contrast between the subconjunctival tissue and sclera. With the $k$-dependence correction but without the coherent spatial Gaussian filter, the contrast between the subconjunctival tissue and sclera was improved, yet fine speckle noise obscured fine structures in the sclera as shown in Fig. 4(c3). The speckle noise in Fig. 4(c3) was caused by the calculation method of local birefringence shown in Eq. (19), where the multiplication of two pixels could enhance the speckle noise. By applying both processing methods, the best image quality of the local retardation was obtained as shown in Fig. 4(c4).

The major structures near the angle were indicated in Fig. 5(a) and Fig. 5(b) that were enlarged images of Fig. 4(b) and Fig. 4(c4) at the light-blue boxed region in Fig. 4(b). The sclera had unevenly high local retardation. The region near the scleral spur and the Schwalbe’s line including the trabecular meshwork had high local retardation, which would be useful for diagnosis of the angle-closure glaucoma as suggested previously [18,45]. Iris pigment epithelium, ciliary body and choroid indicated by white arrows in Fig. 4(c4) appeared to have high local retardation because of polarization scrambling of melanin [16,17,46]. Corneal stroma that was closely perpendicular to the probing beam appeared to have high local retardation as indicated by a light-blue arrow in Fig. 4(c4), which was also reported previously [21].
The coherent spatial Gaussian filter has an additional benefit at the boundary between the tissue and void regions. Figure 6 shows magnified images at the red and green boxed regions of Fig. 4(b). In the local retardation images without the coherent spatial Gaussian filter, the local retardation was artificially high at the surface of the sample indicated by a light-blue arrow of Fig. 6(c) and at the boundary between the tissue and void space indicated by the white arrows of Fig. 6(c) and Fig. 6(d). These artifacts were caused by the calculation method of the local birefringence using separated pixels [43]. They were mitigated by the coherent spatial Gaussian filter as shown in Fig. 6(e) and Fig. 6(f). Since this filter has the effect of extrapolating the tissue signal to the void region, the resultant local retardation becomes low at the boundary.
Fig. 5. Magnified images of OCT intensity (a) and local retardation with \( k \)-dependence correction and coherent spatial Gaussian filter (b) from the light-blue boxed region in Fig. 4(b). Scale bars show 400 \( \mu \)m in air.

Fig. 6. Magnified images of OCT intensity (a), (b), local retardation with \( k \)-dependence correction without coherent spatial Gaussian filter (c), (d) and local retardation with \( k \)-dependence correction and coherent spatial Gaussian filter (e), (f) from the red and green boxed regions in Fig. 4(b), respectively.
4.2 Measurements of known samples

To validate the processing methods further, we measured 2% intralipid and a quarter waveplate, which have known values of birefringence.

Figure 7 shows the images of the 2% intralipid with or without applying the processing methods. As Fig. 4(c3), Fig. 7(b3) had remaining speckle noise caused by the calculation method of local birefringence shown in Eq. (19). The coherent spatial Gaussian filter was effective to reduce it as shown in Fig. 7(b4). To show the statistical distribution of the images in Fig. 7, the histograms of the local retardation images were plotted in Fig. 8. It was clear that the measured local retardation became closer to zero in Fig. 8(d).

Fig. 7. OCT intensity (a) and local retardation (b) images of 2% intralipid. The local retardation images are tabulated in (b) depending on enabled processing methods as well as Fig. 4. The scale bars show 1 mm in air. The red rectangular region in (a) is used in Fig. 8.
Fig. 8. Histograms of the local retardation images shown in Fig. 7. The histograms of (a), (b), (c) and (d) were plotted from Fig. 7(b1), 7(b2), 7(b3) and 7(b4) in the red region shown in Fig. 7(a), respectively.

A quarter waveplate (WPQ05M-1310, Thorlabs, NJ, USA) was measured as a birefringent sample. For this measurement, the lateral scan was not applied. In the processing, the coherent spatial Gaussian filter was disabled, because the sample does not have the speckle noise. A phase retardation from the front surface to the back surface of the sample and relative orientation of the optic axis were calculated without or with the $k$-dependence correction, and were plotted in Fig. 9. Each plot was calculated from a single A-scan. The orientation was left to be wrapped at 90 degrees. In both of Fig. 9(a) and Fig. 9(b), the results with the $k$-dependence correction were closer to the theoretical values, demonstrating the efficacy of the correction.

Fig. 9. Measured phase retardation (a) and orientation (b) of the quarter waveplate. Red and blue plots indicate the results without and with the $k$-dependence correction, respectively.

4.3 Functioning and nonfunctioning filtering blebs

Trabeculectomy is an ophthalmic surgery to reduce the intraocular pressure (IOP) of the glaucomatous eye by creating a filtering bleb that drains the aqueous flow. Figure 10(a), 10(c) and 10(e) are images of a functioning filtering bleb in a 78-year-old female whose IOP was
10 mmHg 10 days after trabeculectomy on the left eye for primary open angle glaucoma (POAG). The OCT intensity image in Fig. 10(c) shows a large cleft in the filtering bleb. The local retardation image in Fig. 10(e) shows negligibly low birefringence in the bleb wall. An exception is indicated by a white arrow in Fig. 10(e), which is birefringence of a suture as seen in the color photo of Fig. 10(a). The sclera and scleral flap shows birefringence as well as the sclera of the healthy eye. Figure 10(b), 10(d) and 10(f) show images of a nonfunctioning filtering bleb in a 53-year-old male whose IOP was 36 mmHg 5 years after trabeculectomy on the left eye for POAG. The OCT intensity image in Fig. 10(d) shows a narrow cleft in the filtering bleb and partial adhesion of the scleral flap to the bleb wall. The local retardation image in Fig. 10(f) shows apparent birefringence encapsulating the cleft near the inner boundary of the bleb wall as indicated by white arrows. This indicates the scar tissue that prevented the drainage of the aqueous flow from the cleft.

![Fig. 10](image_url)

Fig. 10. Color photos (a), (b), OCT intensity (c), (d) and local retardation (e), (f) images of functioning and nonfunctioning filtering blebs, respectively. Red arrows in (a) and (b) indicate B-scan positions. Yellow diamonds in (c) and (d) indicate the scleral flap created by trabeculectomy. The scale bars show 1 mm in air.

### 4.4 Bleb reconstruction surgery

Bleb reconstruction is an ophthalmic surgery to reopen the drainage path of the aqueous flow by removing the scar tissue and reconstructing the filtering bleb. A filtering bleb that had
decreased functioning in the right eye of an 85-year-old female, which was diagnosed as exfoliation glaucoma and had an IOP of 22 mmHg, was measured by PS-OCT before and after the bleb reconstruction surgery. Figure 11(a), 11(c) and 11(e) show images of the filtering bleb before the surgery. The OCT intensity image in Fig. 11(c) showed narrow cleft of the filtering bleb and partial adhesion of the scleral flap to the bleb wall at the location indicated by a red arrow. In the local retardation image of Fig. 11(e), highly birefringent scar tissue was found at the location of the adhesion indicated by a white arrow. Although the cleft was narrow, it was not fully covered by the scar tissue, which would explain the moderately high IOP of 22 mmHg. Two weeks after the bleb reconstruction surgery for this patient, the IOP decreased to 4 mmHg. A color photo and PS-OCT images after the surgery are shown in Fig. 11(b), 11(d) and 11(f). In the OCT intensity and local retardation images of Fig. 11(d) and 11(f), it was observed that the scar tissue was extracted as indicated by a white arrow and the region of the cleft was enlarged. It is noted that the location of the scar tissue cannot be distinguished from that of the bleb wall without scarring if only the OCT intensity images are available. Local retardation images are invaluable to identify scar tissue objectively.

Fig. 11. Color photos (a), (b), OCT intensity (c), (d) and local retardation (e), (f) B-scan images of a filtering bleb before and after bleb reconstruction surgery, respectively. Red arrows in (a) and (b) indicate B-scan positions. Yellow diamonds in (c) and (d) indicate the scleral flap created by trabeculectomy. The red arrow in (c) and the white arrow in (e) indicate the scar tissue on the filtering bleb wall. White arrows in (d) and (f) indicate the location of the removed scar tissue by bleb reconstruction surgery.
Volume-rendered video clips of the local retardation before and after the bleb reconstruction surgery for the same patient of Fig. 11 are shown in Fig. 12 and Fig. 13, respectively. The movies were created using Voreen 4.4 (voreen.uni-muenster.de) [47]. In these movies, the cleft of the filtering bleb and the scleral flap were manually segmented and visualized with light-blue and red colors, respectively. Before the surgery shown in Fig. 12, the scar tissue covered a large area above the scleral flap, and the cleft region was smaller than the scleral flap. After the surgery shown in Fig. 13, a large part of the scar tissue was removed, and the cleft region was enlarged, indicating successful treatment for this filtering bleb.

Fig. 12. Volume-rendered image of the local retardation before the bleb reconstruction surgery (Media 1).

Fig. 13. Volume-rendered image of the local retardation after the bleb reconstruction surgery (Media 2).

5. Discussion

We developed a new method that generates the $k$-clock by applying the optical quadrature modulation to double the clock frequency as described in Section 2.1. In contrast to a previous approach using an electric 90-degree phase shifter [34], the optical phase shift enabled robust phase shifting that is independent of sweep nonlinearity of the light source. Since the sweep tuning speed of the light source was tuned with a nonlinear curve that was close to sinusoid, the clock had a broad frequency range that could easily superpose on its coherence revival, which typically appears in the swept-source external cavity tunable lasers [48]. The light source used in our system showed the coherence revival at integer multiples of 55 mm. The $k$-clock doubling method was effective in generating a high-frequency clock while avoiding this issue.
We showed that the $k$-dependence correction of the measured Jones matrices could reduce the polarimetric noise effectively. It has both a benefit and limitation compared to the previously published method by Braaf et al. [37] The benefit is that our $k$-dependence correction requires only signals from the sample's surface and does not require signals in or beneath the birefringent tissue. This is preferred in practice for our application, because the anterior eye segment has more variety in structure than the retina. On the other hand, the limitation of our method is the assumption that $J_{in}$ and $J'_{out}$ are $k$-independent. In general, any fiber-optic components are more or less $k$-dependent. However, as long as the $k$-dependence is sufficiently small, the polarimetric noise due to it is mitigated by the coherent spatial Gaussian filter. We note that Villiger et al. developed another approach by mitigating the PMD instead of correcting it [39]. Depending on the system and application, optimal solutions may vary.

The measurement results of the intralipid and the waveplate in Section 4.3 demonstrated the efficacy of our processing methods. Ideally, the local retardation of the intralipid is zero. The difference from the zero retardation in Fig. 7(b4) would be likely due to finite SNR and multiple scattering of the dense scatterers. Although any $k$-dependence that does not satisfy the assumption in Section 3.3 can also cause the polarimetric speckle noise, we did not observe particular indication of it in our experiments. The results of the waveplate measurement were close to the theoretical values as shown in Fig. 9. If $J_{in}$ and $J'_{out}$ are $k$-dependent, results of birefringent samples would become different from their theoretical values. Since the difference was not large in the result of the waveplate measurement, our system would have minor $k$-dependence of $J_{in}$ and $J'_{out}$.

In Section 3.1, we applied the numerical compensation to remove the jitter in the measured spectra using the peak signals generated by the FBGs. We note that the PMD of the system does not influence the peak signals generated by the FBGs in practice. This is because the time delay due to the PMD in the reference arm and the sampling interval of our system are different by several orders of magnitude; the former is in the order of subpicosecond, but the latter is in the order of nanosecond.

In Section 3.3, we showed the method of the $k$-dependence correction. It would be an interesting topic how the noise affects it. Since all elements of the Jones matrix were used multiplicatively in Eqs. (7) and (8), it would be safe to align the system so that none of the matrix elements has no signal at the sample’s surface. Although the noise in Eqs. (11) and (12) was reduced by the averaging process shown in Eq. (5), further studies would be helpful to understand the noise characteristics in PS-OCT.

In principle, the $k$-dependence correction would also be able to compensate a mismatch of chromatic dispersion among matrix elements as a part of the PMD of the system as far as the assumption that $J_{in}$ and $J'_{out}$ are $k$-independent is valid. If optical materials that have polarization-dependent chromatic dispersion are used in the interferometer, one may need to consider their influence. In the case of our system, we did not apply chromatic dispersion compensation in the processing, and we did not observe noticeable polarization-dependent chromatic dispersion in our system.

In Section 3.4, we explicitly showed that Jones matrices can be averaged even if they have cumulative phase retardation. To enable averaging the Jones matrices, we proposed a new method of the global phase cancellation in Eq. (18). Using Eq. (18), one can calculate relative global phase from all elements of the Jones matrices by self-cancelling the phase retardation of the birefringence. Although we utilized it only for the coherent spatial Gaussian filter, it would also be applicable for any purposes that require phase-sensitive measurement, as Ju et al. and Li et al. suggested calculating Doppler phase shift using their equations [42,49].

Compared to the previous methods of the global phase estimation [21,37,42], another characteristic of our global phase estimation shown in Eq. (18) is that no specific element of the Jones matrix is used as a reference of the phase estimation. To compare the performance of the global phase estimation for the spatial filtering of Jones matrices, some previous methods of the global phase estimation were applied to the same data of Fig. 4, and the
resultant local retardation images were shown in Fig. 14. Interestingly, there were no major qualitative differences among these methods in both rectangular and Gaussian window functions. Hence, these methods would work similarly for the purpose of spatial filtering except extreme cases, such as highly unbalanced SNRs of the matrix elements. Figure 14 also suggests that a deviation of the experimental data from the assumption that Eq. (17) should be unitary does not cause unstable results.

![Figure 14](image)

Fig. 14. Local retardation images of the healthy human eye using the global phase estimation methods of Lim et al. [21] (a), (b), Ju et al. [42] and Braaf et al. [37] (c), (d) and Eq. (18) (e), (f), respectively. Rectangular spatial window function was used in (a), (c) and (e), and the Gaussian window function was used in (b), (d) and (f). Figure (f) is same as Fig. 4(c4). A kernel size of the rectangular window was 3 pixels (20.6 µm in tissue) in axial direction and 7 pixels (105 µm) in lateral direction. The kernel size of the Gaussian window was same as Section 3.4. The scale bars show 1 mm in air.

We developed PS-OCT for the anterior eye segment, and demonstrated in vivo measurements of the filtering blebs after trabeculectomy. Our PS-OCT could measure high-quality local retardation images of filtering blebs with the techniques of $k$-clock doubling, $k$-dependence correction and coherent spatial Gaussian filter. The patient with good control of IOP showed low birefringence in the bleb wall, and the patients with poor control of IOP showed high birefringence of scar tissue in the bleb wall. These results were consistent with previous studies [21,22]. Furthermore, we demonstrated PS-OCT imaging of the filtering bleb before and after bleb reconstruction surgery. These results suggested that PS-OCT would be able to help surgeons plan needling revision or reconstruction surgery of the filtering bleb.