Parallel detection of Jones-matrix elements in polarization-sensitive optical coherence tomography

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Abstract: The polarization properties of a sample can be characterized using a Jones matrix. To measure the Jones matrix without assumptions of the sample, two different incident states of polarization are usually used. This requirement often causes certain drawbacks in polarization-sensitive optical coherence tomography (PS-OCT), e.g., a decrease in the effective A-scan rate or axial depth range, if a multiplexing scheme is used. Because both the A-scan rate and axial depth range are important for clinical applications, including the imaging of an anterior eye segment, a new PS-OCT method that does not have these drawbacks is desired. Here, we present a parallel-detection approach that maintains the same A-scan rate and axial measurement range as conventional OCT. The interferometer consists of fiber-optic components, most of which are polarization-maintaining components with fast-axis blocking free from polarization management. When a parallel detection is implemented using swept-source OCT (SS-OCT), synchronization between the A-scans and synchronization between the detection channels have critical effects on the Jones-matrix measurement. Because it is difficult to achieve perfect synchronization using only hardware, we developed a solution using a numerical correction with signals from a static mirror. Using the developed system, we demonstrate the imaging of an anterior eye segment from the cornea to the back surface of the crystalline lens.

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1. Introduction

Anterior-segment optical coherence tomography (AS-OCT) can be used to measure cross-sectional images of an anterior eye segment [1,2]. Recent advancements of wavelength swept lasers for swept-source optical coherence tomography (SS-OCT) have enabled a long axial measurement range from the cornea to the back surface of the crystalline lens [3,4], which is necessary for a morphometric and biometric analysis of an anterior eye segment [5]. Because an image contrast of standard AS-OCT relies on the signal intensity of the light scattering, it is difficult to determine the tissue properties that cannot be structurally resolved in an intensity image. Polarization-sensitive optical coherence tomography (PS-OCT) can provide additional contrast to AS-OCT, e.g., the birefringence of fibrous tissue and polarization scrambling, or the depolarization of melanin [6]. Several technical branches of PS-OCT have been developed. Among them, the Jones matrix approach of PS-OCT acquires all elements of the Jones matrix, which fully characterizes the polarization properties of the target in a coherent manner. This is useful for correcting polarization-dependent signal distortion [7-9] and statistically estimating the polarization properties [10]. However, a nontrivial disadvantage would be the complexity of the signal detection. In particular, for the Jones
matrix measurement, polarization-sensitive responses need to be detected from the target using two different incident polarizations.

To illuminate and distinguish two incident lights with different polarizations for the Jones matrix measurement, early PS-OCT systems used a multiplexing of the amplitude modulations [11] or alternate A-scan encoding [12]. Thereafter, various methods were developed not only for time-domain OCT but also for spectral-domain OCT and SS-OCT [6]. Typical approaches used to detect all elements of the Jones matrix in PS-OCT are summarized in Table 1, which focuses on PS-OCT with SS-OCT implementation for simplicity. We also include the approaches of SS-OCT that measures Stokes vectors because of high similarity between the measurements of Jones matrix and Stokes vectors. An alternate A-scan encoding can also be implemented in SS-OCT (Approach A) [13,14]. This has a drawback in that the effective A-scan rate is reduced in half. A frequency modulation or frequency multiplexing technique (Approach B) requires a resonant device, such as a resonant EOM [15] or a frequency shifter [7,16], and encodes the signal in the frequency domain. Although the effective A-scan rate has no drawback in Approach B, the measurable depth range decreases by half. This is because the depth range is split for different modulation frequencies. The signal-to-noise ratio (SNR) penalty of Approach B depends on the efficiency of multiplexing using the resonant EOM or the frequency shifter. It may also be affected by frequency-dependent roll-off of the detector and other electric components. With depth encoding (Approach C), the signals are simply encoded in the axial depth using a polarization-dependent delay unit before illuminating the sample [8,17,18]. This approach also has a drawback in terms of the depth range. The SNR penalty of Approach C mainly depends on the signal roll-off that originates from the finite coherence length of the swept laser. In the case of interleaved sampling (Approach D), two orthogonally polarized incident lights are combined as interleaved signals using the frequency shifter or frequency combs [19,20]. Although Approach D has no drawback regarding the effective A-scan rate and SNR, the depth range decreases by quarter or half because of the interleaved sampling of the spectral interference.

| Approach | Required hardware | Effective A-scan rate | Depth range | SNR penalty |
|----------|-------------------|----------------------|-------------|-------------|
| A: Alternate A-scan encoding | Non-resonant EOM | Half | No drawback | No drawback |
| B: Frequency multiplexing | Resonant EOM /frequency shifter | No drawback | Half | Dependent on efficiency of multiplexing |
| C: Depth encoding | Polarization-dependent delay unit | No drawback | Half | Dependent on signal roll-off |
| D: Interleaved sampling | Frequency shifter or Fabry-Perot etalons | No drawback | Quarter or half | No drawback |
| E: Parallel detection | Specially designed interferometer | No drawback | No drawback | –3 dB |

Table 1. Typical approaches used to measure Jones matrix in PS-OCT

For imaging an anterior eye segment, both the effective A-scan rate and the depth range are important and should not be degraded in PS-OCT when compared to conventional OCT. However, Approaches A, B, C, and D all have one of these issues, as summarized in Table 1. In this paper, we suggest a new approach to overcome these issues, which is listed as Approach E in Table 1. In Approach E, all elements of the Jones matrix are detected in a parallel manner without encoding the signals in successive A-scans or the signal frequency. We call this parallel-detection PS-OCT (PD-PS-OCT). We describe a detailed method enabling a parallel detection and demonstrate the imaging of an anterior eye segment.
2. Methods

Design of the interferometer

Because our approach to parallel detection requires a complicated configuration of the interferometer, we first describe simplified models used to measure two interferometeric signals associated with two orthogonally polarized incident lights.

Simplified models

Figure 1(a) shows a schematic of the depth-encoding of two incident polarizations. Wavelength-swept light is split into reference and sample arms. In the sample arm, the light is further split and passes through optical fibers with different path lengths, which are labeled H and V. The path of V is longer than that of H because of an additional fiber, as shown in the figure. The lights from H and V are combined at the polarization beam splitter/combiner (PBSC) and are regarded as horizontal and vertical polarizations. The combined light is directed to a sample, backscattered from it, recoupled to the fiber, interfered with by a reference light at a coupler, and detected by a balanced photoreceiver (BPD). Figure 1(b) shows the depth-encoded signals in the signal frequency domain. The H and V signals have a depth separation because of the different path lengths before illuminating the sample. Because the depth range from DC to the Nyquist frequency is divided for the H and V signals, the measurable depth range of the sample is half that of a conventional SS-OCT.

Figure 2(a) shows a simplified model of the parallel detection. The sample arm is mostly same, as shown in Fig. 1(a), although the light recoupled after illuminating the sample is split using a fiber coupler before being interfered with by a reference light, as shown in Fig. 2(a). The reference light is also split to apply the same difference in path length between H and V of the sample arm, as shown in Fig. 2(a). These signals are individually detected by two BPDs, which are shown in Fig. 2(b). For Ch. 1 of Fig. 2(b), signal H is within a measurable depth range from DC to the Nyquist frequency, whereas signal V is outside this range and the aliasing of signal V is suppressed by the finite bandwidth of the BPDs and electric low-pass filters, which are not shown in Fig. 2. For Ch. 2 of Fig. 2(b), signal V is within the
measurable depth range, and the conjugate of signal H is outside this range and is suppressed. Unlike the depth-encoding shown in Fig. 1, parallel detection can use the full depth range from DC to the Nyquist frequency for imaging.

(a)

![Parallel Detection Schematic](image)

(b)

![Depth Separation](image)

Fig. 2. Schematic of parallel detection for interferometric signals of two incident polarizations: (a) fiber-based interferometer and (b) signals detected by two channels. The same abbreviations as in Fig. 1 are used. The two detection channels, Ch. 1 and Ch. 2, detect the signals of H and V, respectively.

Implementation of PD-PS-OCT

Although the simplified models in Figs. 1 and 2 describe how to encode the two orthogonally polarized incident lights, and omit a polarization analyzer for detecting the state of polarization after illuminating the sample, both features are necessary to detect all $2 \times 2$ elements of the Jones matrix. Figure 3 shows a schematic of the implemented PD-PS-OCT for measuring the Jones matrix using the parallel detection and polarization-sensitive detection of the backscattered light from the sample. In this subsection, we describe the detailed configuration of the PD-PS-OCT in a step-by-step manner.

The light source is a short-cavity wavelength-strept laser (Axsun Technologies, Billerica, MA, US) with a center wavelength of 1,304 nm, a wavelength tuning range of 106 nm, a sweep rate of 50 kHz, and an output power of 23.2 mW. The light is directed to a polarization-maintaining filter coupler (PMFC, Optizone Technology Ltd., Shenzhen, China) that has a single-mode fiber (SMF) as input port, polarization-maintaining fibers (PMF) as output ports, and a beamsplitter with fast-axis blocking of the PMF. This PMFC is labeled as PMFC1. All PMFCs are fast-axis blocked in this system with a minimum extinction ratio of 20 dB. In Fig. 3, the black and colored (blue, red, orange, green, and magenta) curves indicate SMF and PMF, respectively. Note that the colors of the PMFs in the figure are applied for clarification, but do not indicate different types of fibers. A 10% portion of the light split by the PMFC is directed to a module including an auxiliary interferometer (AUX-I) and fiber Bragg grating (FBG) to output a sampling clock that is linear to the wavenumber, which we
call the \( k \)-clock, and a sampling trigger. AUX-I is configured as a quadrature modulator, and the signal frequency is doubled to \( \sim 450 \) MHz using a logical circuit [8]. The other 90% of the light is directed to the second PMFC (PMFC2), which splits the light into the sample and reference arms at a ratio of 80/20.

![Fig. 3. Configuration of the PD-PS-OCT. Abbreviations are as follows: WSL, wavelength-swept laser; PC, polarization controller; PMFC, polarization-maintaining fiber coupler; AUX-I, auxiliary interferometer; FBG, fiber Bragg grating; PMF, polarization-maintaining fiber; PBSC, polarization beam splitter/combiner; SMFC, single-mode fused coupler; GB, glass block; FR, Faraday rotator; VDL, variable optical delay line; HSD, high-speed digitizer; PD, electric power divider.](image)

In the sample arm, which is shown in the upper-right part of Fig. 3, the PMFC3 further splits the light with a ratio of 50/50. One of the output ports has a longer (plus one meter) PMF than the other port. The light passes through a polarization-maintaining circulator (PMCIR, Optizone Technology Ltd., Shenzhen, China) in each port. The PMCIRs block the light in the fast axis of the PMFs for all ports with a minimum extinction ratio of 20 dB. The two paths of the sample arm are combined at a polarizing beam splitter/combiner (PBSC1) as orthogonal polarizations. The output port of the PSBC is an SMF and is connected to a single-mode fused coupler (SMFC). The SMFC splits the light toward a sample and a static mirror with a ratio of 99/1. The static mirror, referred as calibration mirror, is used to correct the signals, as described in a later section. Before illuminating the calibration mirror, a quarter of the cross-sectional area of the collimated beam is covered by a glass block (GB) to apply two different path lengths; one path length is generated by a portion of the light that passes only once through the GB, whereas the other is generated by the portion of light that does not traverse the GB. We note that a third path length is possible as the light may pass twice through the GB. The cross-sectional area of the calibration mirror is aligned to locate the two signals from the calibration mirror near the Nyquist frequency. The resultant depths of the two signals are 0.89 and 0.95, where the depth range from DC to the Nyquist frequency is normalized to be 0 to 1. In addition, a polarization controller (PC) in the fiber path toward the calibration mirror is aligned such that the return light from the calibration mirror is almost equally coupled to the two PMFs at the PBSC1. All calibration signals have
SNRs of ~30 dB. At the 99% port of the SMFC, the light is collimated, scanned by two
galvanometer scanner mirrors, and focused toward the anterior eye segment by an objective
lens. The probe power is 9.24 mW. Backscattered light from the eye and light reflected from
the calibration mirror are coupled to the SMFs and directed toward the PBSC1. At the
PBSC1, each orthogonal component of the polarization is split and directed to the PMCIR
(PMCIR1 or PMCIR2) and a 50/50 PMFC (PMFC4 or PMPC5) along each path. Each port is
connected to another 50/50 PMFC (PMFC6-9) to interfere with the lights from the reference
arms.

In the reference arm, which is shown in the lower-middle part of Fig. 3, the light passes
through a PBSC2, PMF, and SMF, and is directed to a common variable delay line (VDL,
General Photonics, Chino, CA, US). The connection between the PMF and the SMF has no
significant coupling loss because of similar mode field diameters. The common VDL includes
a Faraday rotator (FR) to rotate the state of polarization of the reflected light for 90° at the
PBSC. Because this FR is common to all signals, it does not cause wavelength-dependent
polarization mode dispersion. The reflected light from the common VDL passes through the
PBSC2 and is directed to a 50/50 PMFC10. One of the output ports has a 1-m longer PMF,
similar to the sample arm. Each port is further split using a 50/50 PMFC (PMFC11 or
PMFC12). All four output ports are connected to individual VDLs to adjust the differences in
path length caused by the tolerance of the fiber lengths in the sample and reference arms,
except for the extension of the 1-m PMFs. The individual VDLs are folded using
retroreflectors, which is not shown in Fig. 3, and have separate input and output ports. Each
output of the individual VDL is directed to the 50/50 PMFC (PMFC6-9) to interfere with the
light from the sample arm.

The light interference is detected using a balanced photoreceiver (BPD; C12668-02,
Hamamatsu Photonics, Hamamatsu, Shizuoka, Japan) in each channel. The signals from the
four detection channels are filtered using low-pass filters (SLP-250 +, Mini-Circuits,
Brooklyn, NY, US) and acquired using two dual-channel high-speed digitizers (HSDs; APX-
5200A, Aval Data Corporation, Machida, Tokyo, Japan), which have 12-bit resolution, a
maximum sampling rate of 1 GSamples/s, and ~3 dB analog bandwidth of 375 MHz. To
match the signals arriving at the HSDs with the k-clocks and triggers, 1-m long coaxial cables
are added for the k-clock and the trigger of HSD1, as shown in Fig. 3. After applying the
signal processing described in the following, the system sensitivity and axial resolution of the
OCT intensity are 101 dB and 10.6 µm in tissue independent of the four detection channels,
respectively.

**Signal processing**

Ideally, all detection channels in Fig. 3 should be synchronized to the wavelength sweep of
the light source through the trigger and k-clock. In practice, however, it is known that the
synchronization between the detection channels and the synchronization between the repeated
wavelength sweeps can experience fluctuations. Similar issues that are potentially caused by a
collision of the trigger and k-clock, as well as by individual differences in the analog-to-
digital converters depending on the channels, have been previously reported [8,21,22]. It is
important for PD-PS-OCT to correct the synchronization error between all four detection
channels and between all A-scans. In this study, we developed a new method that fits our
system to compensate the fluctuation using the signals from the calibration mirror. Herein, we
describe the processing flow that begins from the measured raw spectra.

**Preprocessing**

In each B-scan and detection channel, a DC subtraction is applied to the measured raw spectra
by subtracting an average low-pass filtered spectrum. For all detection channels, a wavelength
dispersion mismatch between the reference and sample arms are corrected with a second-
order correction function of the spectral phase optimized for one of four channels. The spectra
are converted into an axial depth domain using a discrete Fourier transform (DFT). Because the individual VDLs of the reference arm cannot be aligned to match the path lengths of all channels perfectly, the differences in residual axial depth of the calibration signals between the detection channels are numerically corrected by shifting the signals in the axial depth domain within a multiple integer accuracy of the axial depth pixels. In other words, the four A-scan signals are aligned with single-pixel accuracy. At this processing stage, the signals have differences in the subpixel axial depth and a residual dispersion mismatch between the four channels. The latter is because the lengths of the PMFs have tolerance in the interferometer, and the PMF (SM13-PS-U25A, Fujikura Ltd., Koto-ku, Tokyo, Japan) is dispersive unlike the SMF at 1,310 nm (SMF-28e, Corning Inc., NY, US). Within the spectral domain, both the former and latter can be corrected numerically using first- and second-order corrections of the spectral phase. For this purpose, one of two signals from the calibration mirror in the axial depth domain is extracted using a rectangular window function with a width of 7 pixels (67 µm), converted into the spectral domain using an inverse DFT for each detection channel, and averaged as

\[
E'_{j, \text{mirror}}(k) = \frac{1}{n_{\text{all}}} \sum_{n=1}^{n_{\text{all}}} E_{j, \text{mirror}}(k, n) \exp\left\{-i \arg(E_{1, \text{mirror}}(k, n))\right\}, \tag{1}
\]

where \( E_{j, \text{mirror}}(k, n) \) is the complex interferometric spectrum of the calibration mirror in the spectral or wavenumber \( k \) domain at the \( n \)-th wavelength scan (A-scan) of the \( j \)-th detection channel, \( n_{\text{all}} \) is the number of wavelength scans in a single lateral scan of the sample (B-scan), and \( i \) is the imaginary unit. In this paper, \( n_{\text{all}} \) is 800 for the waveplate measurement and 1,600 for the imaging of the anterior eye segment, whose results are shown in Section 3. As defined in Eq. (1), \( E'_{j, \text{mirror}}(k) \) is the averaged spectrum normalized by an argument of \( E_{1, \text{mirror}}(k,n) \). We apply phase unwrapping and second-order polynomial fitting to

\[
\arg(E'_{j, \text{mirror}}(k))
\]

along \( k \), and define the resultant spectral phase as \( \theta_j(k) \). The complex interferometric signal without extracting the calibration signal, \( E_j(k,n) \), is then corrected using \( \theta_j(k) \) as

\[
E'_j(k,n) = E_j(k,n) \exp\left\{-i\theta_j(k)\right\}. \tag{2}
\]

Applying DFT to Eq. (2), we calculate a complex signal in the axial depth \( z \) as

\[
E'_j(z,n) = \mathcal{F}\left[E'_j(k,n)\right], \tag{3}
\]

where \( \mathcal{F}[-] \) indicates the DFT.

Detection of unsigned integer-multiple shift in measured spectra between A-scans

In the second step of the signal processing, we detect an unsigned integer shift in the measured spectra between A-scans. The sign of the shift is separately determined through a successive process for the robustness, as described later.

We define \( E'_j(z,n) \equiv E'_j(z,n) \) as a temporary variable. If the dependency on the \((n-1)\)-th and \( n \)-th A-scans is represented as \( \Delta_{n-1,n} \), the relative phase between the \((n-1)\)-th and \( n \)-th A-scans at the \( j \)-th detection channel is

\[
\arg\left[E'_j(z,n)E''_j(z,n-1)\right] = \eta_j(z,\Delta_{n-1,n}) + \varphi_j(\Delta_{n-1,n}) + \xi_j(\Delta_{n-1,n}), \tag{4}
\]
where the superscript asterisk shows a complex conjugate, \( \eta_j(z, \Delta_{n-1,a}) \) is a linear phase shift along depth \( z \) caused by the integer shift of data acquisition between the \( (n-1) \)-th and \( n \)-th A-scans, \( \varphi_j(\Delta_{n-1,a}) \) is a phase fluctuation of the calibration signal caused by a path length fluctuation of the interferometer between the \( (n-1) \)-th and \( n \)-th A-scans, and \( \xi_j(\Delta_{n-1,a}) \) is the relative phase noise because of a finite SNR of the calibration signal [23]. Assuming that the detected spectra have only a \( \pm 1 \) pixel fluctuation of the data sampling between the successive A-scans, the linear phase shift \( \eta_j(z, \Delta_{n-1,a}) \) in Eq. (4) can be theoretically described from the shift theorem of a Fourier transform [24] as

\[
\eta_{\text{theory}}(z_{m1}) = \pm \frac{z_{m1}}{z_{\text{max}}} \pi, \tag{5}
\]

where \( z_{m1} \) and \( z_{\text{max}} \) show the axial depths of the calibration mirror and Nyquist frequency, respectively. Consequently, the shift in the detected spectra can be determined within \( \pm 1 \) pixel using Eq. (5). In principle, we can correct the linear phase shift along the axial depth by replacing \( \epsilon_j'(z, n) \) in a repetitive manner for the A-scan number \( n \) as

\[
\begin{align*}
\epsilon_j'(z, n) & 
\begin{cases}
\epsilon_j'(z, n) e^{i \eta_{\text{theory}}(z_{m1})} & \text{if } |\arg[\epsilon_j'(z, n) e^{i \eta_{\text{theory}}(z_{m1})}]| \geq |\eta_{\text{theory}}(z_{m1})|/2 \\
\epsilon_j'(z, n) & \text{otherwise}
\end{cases}

\rightarrow \epsilon_j'(z, n) e^{i \eta_{\text{theory}}(z_{m1})} & \text{if } |\arg[\epsilon_j'(z, n) e^{i \eta_{\text{theory}}(z_{m1})}]| \leq |\eta_{\text{theory}}(z_{m1})|/2 .
\end{align*}
\tag{6}
\]

Equation (6) is repeated for all A-scans from \( n = 2 \) to \( n_{\text{all}} \). If \( z_{m1} \) is close to \( z_{\text{max}} \), however, \( |\arg[\epsilon_j'(z, n) e^{i \eta_{\text{theory}}(z_{m1})}]| \) can be aliased because of \( \varphi_j(\Delta_{n-1,a}) \) and \( \xi_j(\Delta_{n-1,a}) \), and the sign of the correction may be inaccurate. We therefore use Eq. (6) only to determine an absolute (unsigned) shift of the detected spectra \( f_j^{\text{abs}}(\Delta_{n-1,a}) \) as

\[
f_j^{\text{abs}}(\Delta_{n-1,a}) = \begin{cases}
1 & \text{if } |\arg[\epsilon_j'(z, n) e^{i \eta_{\text{theory}}(z_{m1})}]| \geq |\eta_{\text{theory}}(z_{m1})|/2 \\
0 & \text{otherwise}
\end{cases}, \tag{7}
\]

and leave the resolution to the sign ambiguity for the next step.

A numerical example of the calibration mirror at \( z_{m1} \) is shown in Fig. 4. Figure 4(a) shows plots of \( E_j'(z_{m1}, n) \) in Eq. (3), where \( \eta_{\text{theory}}(z_{m1}) \) is \( \pm 2.837 \) rad. Abrupt phase changes along the A-scan bins in Fig. 4(a) are detected using Eq. (7), and are shown in Fig. 4(b). In this example, only the channels of \( j = 1 \) and \( 2 \) have multiple-integer shifts in the measured spectra between A-scans.
Detection and correction of signed integer-multiple shift in measured spectra

In the previous subsection, we derived an unsigned integer shift between A-scans in the measured spectra using one of two calibration signals at depth $z_{m1}$ for each detection channel $j$. In this subsection, we derive the sign of the integer shift between A-scans in the measured spectra together with a signed integer shift between detection channels. Complex signals of two calibration signals at depths $z_{m1}$ and $z_{m2}$ in the axial depth domain can be rearranged from Eq. (3) as

$$E_j'(z_{m1}, n) = e^{i\vartheta_{m1}} E_j^x(z_{m1}, n),$$

$$E_j'(z_{m2}, n) = e^{i\vartheta_{m2}} E_j^x(z_{m2}, n),$$

where $\vartheta_{m1}$ and $\vartheta_{m2}$ are arbitrary phases, and we assume $z_{m1} > z_{m2}$. We define $\vartheta_{m1}$ and $\vartheta_{m2}$ such that $\arg[E_j'(z_{m1}, n)] = \arg[E_j'(z_{m2}, n)] = 0$ without a loss of generality. We derive a relative complex value between the detection channels 1 and $j$ for a relative complex signal between the two depths $z_{m1}$ and $z_{m2}$ as

$$g_j(n) = \frac{E_j'(z_{m1}, n)}{E_j'(z_{m2}, n)} \left[ \frac{E_j^x(z_{m1}, n)}{E_j^x(z_{m2}, n)} \right]^{-1} = \frac{E_j'(z_{m1}, n)}{E_j'(z_{m2}, n)} \left[ \frac{E_j^x(z_{m1}, n)}{E_j^x(z_{m2}, n)} \right]^{-1}.$$  (10)

We derive the averaged values of $g_j(n)$ depending on $f_j^{\text{abs}}(\Delta_{n-1,n})$ as
Using Eqs. (7) and (11), it is possible to determine and correct the signed integer-multiple shift of measured spectra, $\chi_j(n)$, at the $j$-th detection channel and $n$-th A-scan relative to the first detection channel and first A-scan within a $\pm 1$ pixel shift of the data sampling. Because this algorithm used to determine $\chi_j(n)$ has many conditional branches, it is summarized as a flowchart in Fig. 5. Note that a condition $|\arg(g_j^{\text{abs}0})| \geq |\arg(g_j^{\text{abs}1})|$ is used instead of a condition $|\arg(g_j^{\text{abs}0})| > \frac{\pi}{2} + \frac{2\pi \Delta n}{r_{\text{scan}}}$ to determine whether $g_j^{\text{abs}0}$ is shifted, when $g_j^{\text{abs}1}$ is available in Fig. 5. This process is applied to each B-scan.

Using $\chi_j(n)$, the complex signal $E_j'(z,n)$ in Eq. (3) is corrected as
\[ E_j^r(z,n) = E_j^r(z,n) \exp \left[ -\frac{z}{\max} \pi \cdot \chi_j(n) \right]. \]  \hspace{1cm} (12)

A numerical example of Eq. (10) is shown in Fig. 4(c), where a theoretical value of \([ (z_i - z_{m2})/z_{max} ] \pi \) is 0.177 rad. Because Eq. (10) is relative to the channel \( j = 1 \), other channels of not only \( j = 2 \) but also \( j = 3 \) and 4 have abrupt phase changes, unlike those shown in Fig. 4(a) and (b). After the correction using Eq. (12), \( \arg \left[ E_j^r(z,n) \right] \) is as shown in Fig. 4(d), which includes a channel-dependent phase drift.

**Correction of phase drift**

In the previous subsection, we corrected only the multiple-integer sampling errors, namely, \( \eta_j(z,\Delta_{n-1,n}) \), in Eq. (4). In addition, the phase fluctuation of the calibration signal caused by a path length fluctuation of the interferometer, \( \phi_j(\Delta_{n-1,n}) \) in Eq. (4), should also be corrected. Using the mirror signal at depth \( z_{m1} \), it is implemented as

\[ E_j^r(z,n) = E_j^r(z,n) \frac{E_j^r(z_{m1},n)}{E_j^r(z_{m1},n)}. \]  \hspace{1cm} (13)

**Correction of subpixel shift**

When we corrected the multiple-integer shift of the measured spectra in Eq. (12), we implicitly assumed that there was no subpixel shift in the measured spectra. In practice, however, it is nonzero because of small discrepancies in the lengths of the optical fibers and coaxial cables between the detection channels. It can be assumed that this subpixel shift of the measured spectra is static if we use the same hardware. We therefore measure a static mirror as a sample at several depths \( z_l \), where \( l \) is an index of the depth location from \( l = 1 \) to \( l_{max} \), and \( z_l \) is sparsely distributed to cover the depth range from \( z = 0 \) to \( z_{max} \). A relative phase \( \zeta_j(l) \) of the complex OCT signals between the first and \( j \)-th detection channels at the \( l \)-th depth is derived using \( n_{all} \) A-scans as

\[ \zeta_j(l) = \arg \left\{ \frac{1}{n_{all}} \sum_{n=1}^{n_{all}} \exp \left[ i \cdot \arg \left( \frac{E_j^r(z_l,n)}{E_j^r(z_{m1},n)} \right) \right] \right\}. \]  \hspace{1cm} (14)

A relative phase between \( \zeta_j(1) \) and \( \zeta_j(l) \) is derived as

\[ \zeta_j(l) = \arg \left\{ \exp \left[ i \left( \zeta_j(l) - \zeta_j(1) \right) \right] \right\}, \]  \hspace{1cm} (15)

which is sufficient for calculating the linear phase slopes in the axial depth caused by the subpixel shifts in the measured spectra. We calculate the phase slope \( a_j \) using all \( \zeta_j(l) \) for \( l = 1 \) to \( l_{max} \), and correct the complex signal in Eq. (13) as

\[ E_j^{\prime\prime\prime}(z,n) = E_j^r(z,n) \exp( -i a_j z). \]  \hspace{1cm} (16)

In the case of our implementation, the phase slope \( a_j \) was measured using \( l_{max} = 13 \) and resulted in \( a_2 = -4.95 \times 10^{-5} \), \( a_3 = -3.82 \times 10^{-5} \), and \( a_4 = -4.21 \times 10^{-5} \) rad/\( \mu \)m.

**Rearrangement of Jones matrix**

The phases of the corrected signals in Eq. (16) are relative to the signal of the calibration mirror at the detection channel \( j = 1 \) and depth \( z_{m1} \). Although the Jones matrix of a sample with negligible diattenuation is unitary in principle, Eq. (16) is no longer unitary. When a relationship between the detection channels and elements of the Jones matrix is written as
it can be rearranged into a unitary as

\[
J'(z,n) \equiv \begin{bmatrix}
E_1''(z,n) & E_2''(z,n) \\
E_3''(z,n) & E_4''(z,n)
\end{bmatrix}.
\]

Note that a similar processing of the rearrangement was also previously applied [8,9].

Using the above processing, the correction of the synchronization error and the rearrangement of the Jones matrix are completed.

3. Results

Measurement of waveplate

To validate our methods, we measured a quarter waveplate at 633 nm (WPQ10ME-633, Thorlabs Inc., Newton, NJ, US), which has a double-pass retardation of 1.35 rad at 1,310 nm. The optic axis of the waveplate was rotated from 0° to 180° with 10° steps during the measurements. The differential Jones matrix in the axial depths between the front and back surfaces of the waveplate was eigendecomposed to calculate the eigenvalues and eigenvectors [8]. The double-pass phase retardation was calculated from the eigenvalues. The optic axis of the waveplate was calculated from the angle of the Stokes parameters converted from the eigenvectors. The results of the phase retardation and orientation are plotted in Fig. 6(a) and (b), respectively. In Fig. 6, three different methods of a signal correction were compared, namely, Eq. (6), which uses only a single calibration mirror; Eq. (13), which uses two calibration mirrors; and Eq. (16), which corrects the subpixel shift in addition to the correction of Eq. (13). The plots with a correction using Eq. (6) showed a high discrepancy from the theoretical values and high tolerance depending on the set orientation, indicating an inaccurate correction for the synchronization error. The plots with a correction using Eq. (13) showed a smaller discrepancy from the theoretical values than when using Eq. (6) for both the phase retardation and orientation. However, the phase retardation with correction using Eq. (13) in Fig. 6(a) showed a small sinusoidal modulation depending on the set orientation, which implies a remaining systematic error in the measured Jones matrix. After the correction using Eq. (16), the discrepancy of the phase retardation was minimal with no dependency on the set orientation. The correction using Eq. (16) was also effective regarding the orientation, as shown in Fig. 6(b).
Fig. 6. (a) Plots of phase retardation and (b) orientation of the waveplate using different correction methods. Applying more advanced correction, better agreement with the theoretical values is obtained.

Measurement of human eye in vivo

A healthy right eye of a human volunteer (37-year-old Japanese subject) was measured using the developed system for a demonstration of an anterior eye segment imaging applied in vivo. A single B-scan with a lateral scanning range of 16 mm consisting of 1,600 A-scans was used for the processing. After applying the correction described above, a Cloude–Pottier decomposition was used as an image filter and to calculate the local retardation per unit depth and noise-bias corrected entropy of the Jones matrix [10]. Images of the eye scanned horizontally and centered at the cornea and at an angle of the temporal side are shown in Fig. 7(a-c) and (d-f), respectively. In Fig. 7(a) and (d), the anterior eye segment was successfully imaged from the cornea to the back surface of the crystalline lens. The trabecular meshwork showed high local retardation, as indicated by the white solid arrows in Fig. 7(b). Although the iris pigment epithelium also showed high local retardation, as indicated by the light-blue dashed arrows in Fig. 7(b) and (e), it was an artifact of depolarization by melanin pigment [25], as shown in Fig. 7(c) and (f). The ciliary epithelium and a part of the ciliary body also showed high entropy because of melanin pigments, as indicated by the solid green and dashed arrows in Fig. 7(f), respectively. In Fig. 7(f), it was observed that the high entropy of the ciliary epithelium was continuous from pars plicata at the center of the image including the solid green arrow to pars plana at the left side of the image beneath the sclera. Although the other part of the ciliary body, indicated by an orange arrow in Fig. 7(f), showed low entropy, a further study with a better SNR is required to conclude whether this is reliable. The corneal stroma also demonstrated depolarization only when the cornea was almost perpendicular to the probing beam, as indicated by the light-blue solid arrows in Fig. 7(b), (c), (e), and (f). This result is consistent with previous studies [10,26]. The lateral rectus muscle and the tendon connected to it [27] showed high local retardation, as indicated by the solid blue arrow in Fig. 7(e). The sclera, which continues from the trabecular meshwork to the periphery of the image under the lateral rectus muscle, showed moderate local retardation in Fig. 7(e) and moderate entropy in Fig. 7(f) because of the interwoven structure of the collagen fiber bundles. The crystalline lens shows no noticeable local retardation in Fig. 7(b) and (e).
4. Discussion

Because the state of polarization in SMF can easily change depending on the temperature and fiber bending [31], PMF is a promising alternative to SMF for the building of interferometers used in clinical settings. The use of PMF has been suggested since the early 2000s [32-34]. However, it is difficult to remove ghost signals because of finite extinction ratios at the connections of the PMFs and PMF-based components. Although these ghost signals can be separated by moving them outside the axial imaging range using long PMFs of more than tens of meters in length [35-37], this increases the required cost. In our design of the interferometer shown in Fig. 3, we used only one of two polarization modes in the PMFs to suppress the ghost signals by applying fast-axis-blocked PMF-based components. An exception to this was a part of the interferometer after passing through the PBSC1 in the sample arm. This part was built using SMF-based components, where two polarization modes must be combined and transmit the fiber before illuminating the eye. Initially, we built this part in a free space to maintain the state of polarization through the use of bulky beamsplitters instead of PBSC and SMFC in Fig. 3. However, we rejected this design because speckle patterns of individual OCT channels are too sensitive to the alignment for the matching of two orthogonally polarized beams in a free space from the two PMCIRs, resulting in a noisy Jones matrix.

In our current design shown in Fig. 3, two polarization modes of the backscattered light from the eye was split at the PBSC1 and directed to two PMCIRs in the sample arm. The design was also aimed at suppressing the ghost signals using fast-axis-blocked PMF-based
components. In almost all previous implementations of polarization-diverse detection, two orthogonal components of polarization have been separated in a detection arm, which occurs after interfering with the lights from the sample and reference arms [6,38]. This configuration ensures a phase relationship between the detection channels. In our PD-PS-OCT, however, we intentionally break this condition by separating the two orthogonal components of polarization in the sample arm, as shown in Fig. 3. This enables us to use fast-axis-blocked PMF-based components for the suppression of ghost signals, as mentioned above, which is also important to parallelize the detection of the Jones-matrix elements.

A drawback of PD-PS-OCT is the SNR penalty of $-3$ dB, as shown in Table 1. Although an inevitable loss occurs in PD-PS-OCT, it is within an amount comparable to the recoupling loss of the polarization-diverse detection built in a free space, which has been used by a large number of previous PS-OCT systems [7-9, 16, 20]. We therefore believe that the impact of such loss is minor in both cases.

Because parallel detection has another drawback during synchronization, it is corrected using two calibration signals, as described in Section 2. Many approaches have been previously developed for similar issues. In the early days of phase-sensitive SS-OCT, interferometric signals acquired using an internal clock of HSD were corrected using single or plural signals from a calibration mirror or a Mach–Zehnder interferometer [15, 24, 39-41]. Although it is effective for passive synchronization to use a $k$-clock and trigger based on a gas cell [42] or FBG, a timing collision between the $k$-clock and trigger must be carefully avoided [21]. An error in the passive synchronization can be numerically corrected using FBG [8,22], a calibration mirror [43], or signals of the sample itself [44-46]. To develop our correction method for PD-PS-OCT, we assumed rather tight conditions as follows: multiple FBGs [8] may not have exactly the same property of spectral reflection, a timing collision between the $k$-clock and trigger may occur, and signals of the sample in some of the detection channels may have an insufficient SNR for the correction using the sample itself because of an unknown birefringence of SMFs prior to illuminating the sample. Our proposed correction method described in Section 2 considers and avoids these potential issues.

As described in Section 2, we use two calibration signals that are separated by the GB and positioned near the Nyquist frequency. These two calibration signals can be positioned at the other depths. For example, one of the signals can be positioned near the zero-delay depth to have better SNR of Eq. (10). However, this configuration restricts the depth position of the sample more strictly than our current configuration. If the cornea is measured at the center of the depth range where the focusing is optimal, a portion of the signal from the crystalline lens overlays the calibration signal near the zero-delay depth, resulting in failure of the measurement. In addition, if the eye has floaters in the vitreous body near the crystalline lens, it can also result in the failure of the measurement. On the other hand, in the previously described configuration and in our current configuration of the calibration signals, the eyelash can overlay the calibration signal(s) and easily cause the failure of the measurement. However, this scenario can be ignored, because the signal from the eye is shadowed by the eyelash and cannot be detected regardless of the calibration. Considering the above discussion, we position two calibration signals only near the Nyquist frequency in Section 2.

We implement the signal processing in Section 3 using LabVIEW 2017 (National Instruments, Austin, TX, US), where we run the code mostly on an 8-core CPU (Core i7-6900K, Intel Corporation, Santa Clara, CA, US) except for some DFTs and the Cloude-Pottier decomposition by general-purpose computing on graphics processing units (GPGPU) using GeForce GTX1080 (Nvidia Corporation, Santa Clara, CA, US). In this environment, the processing time is 13 seconds for the single B-scan in Fig. 7. We are currently porting all the processing flow to the GPGPU, and we estimate the processing time to be 1 second or less per B-scan.

In Section 2, we implicitly assume that the measured Jones matrix is a unitary matrix multiplied by a certain common gain factor. In practice, it is not true because of tolerances of
the insertion and coupling losses depending on the optical paths. In our system, the optical powers of the two incident polarized lights at the sample and of the eight reference lights at the four BPDs have a difference of ~4% and differences within ± 11%, respectively. Although we did not correct the amplitudes of the measured matrix elements in Section 2, it does not influence the eigenvalues of the differential Jones matrix [47]. However, it does influence the eigenvectors of the differential Jones matrix in principle. Although the results of the optic axis in Fig. 6(b) agree with the set orientation well, the additional numerical correction to the amplitudes may improve the results of the optic axis, which is our future issue.

The interferometer of the PD-PS-OCT shown in Fig. 3 uses a large number of fiber-optic components compared to previous PS-OCT systems. Although PD-PS-OCT requires additional cost, all components are commercially available without heavy customization. In addition, the issues of complexity and cost in PD-PS-OCT may potentially be improved through the application of a photonic integration circuit, which has already been demonstrated in miniature implementations of conventional OCT [48-51] and depth-encoded PS-OCT [52].

Although OCT imaging within a full depth range from the cornea to the back surface of the crystalline lens in the anterior eye segment has been demonstrated using both experimental [53-55] and commercial devices [4], to the best of our knowledge, this is the first demonstration of the same region using PS-OCT. In addition to various clinical applications of anterior-segment OCT [5], PD-PS-OCT has the potential to provide additional contrast for an anterior eye segment without compromising the range of axial depth.

5. Conclusions

We developed PD-PS-OCT as a novel approach to measuring all elements of a Jones matrix in a parallel manner. With this method, two orthogonal polarizations are multiplexed using a path length difference of longer than the cutoff frequency of the detection. The backscattered light from the sample is directed to four parallelized couplers and interfered with by four reference lights whose path lengths are managed to detect the Jones matrix. An algorithm was developed to correct the synchronization error using two calibration signals between the A-scans and that between the detection channels. The measurement results using a waveplate validate the efficacy of the algorithm. Using PD-PS-OCT, we demonstrated the application of PS-OCT imaging of an anterior eye segment. The OCT intensity, local retardation, and entropy images were shown through a single B-scan by keeping the same axial depth range as in conventional SS-OCT. PD-PS-OCT is a promising approach to the detection of the Jones matrix of an anterior eye segment, and we believe that its capability is not limited to this particular application. In the field of ophthalmic imaging, the proposed approach can also be applied to retinal or whole-eye imaging within the 840 or 1,064 nm wavelength bands, where a probing light can be transmitted to the retina [3,56].

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Disclosures

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References

1. J. A. Izatt, M. R. Hee, E. A. Swanson, C. P. Lin, D. Huang, J. S. Schuman, C. A. Puliafito, and J. G. Fujimoto, “Micrometer-scale resolution imaging of the anterior eye in vivo with optical coherence tomography,” Arch. Ophthal. 112(12), 1584–1589 (1994).

2. R. Steinert and D. Huang, Anterior Segment Optical Coherence Tomography (SLACK, 2008).

3. I. Grulkowski, J. J. Liu, B. Potsaid, V. Jayaraman, C. D. Lu, J. Jiang, A. E. Cable, J. S. Duker, and J. G. Fujimoto, “Retinal, anterior segment and full eye imaging using ultrahigh speed swept source OCT with vertical-cavity surface emitting lasers,” Biomed. Opt. Express 3(11), 2733–2751 (2012).

4. T. Shoji, N. Kato, S. Ishikawa, H. Ibuki, N. Yamada, I. Kimura, and K. Shinoda, “In vivo crystalline lens measurements with novel swept-source optical coherence tomography: an investigation on variability of measurement,” BMJ Open Ophthalmol 1(1), e000058 (2017).

5. M. Ang, M. Baskaran, R. M. Werkmeister, J. Chua, D. Schmid, V. Aranha Dos Santos, G. Garhöfer, J. S. Mehta, and L. Schmetterer, “Anterior segment optical coherence tomography,” Prog. Retin. Eye Res 66, 132–156 (2018).

6. J. F. de Boer, C. K. Hitzenberger, and Y. Yasuno, “Polarization sensitive optical coherence tomography - a review [Invited],” Biomed. Opt. Express 8(3), 1838–1873 (2017).

7. E. Z. Zhang, W.-Y. Oh, M. L. Villiger, L. Chen, B. E. Bouma, and B. J. Vakoc, “Numerical compensation of system polarization mode dispersion in polarization-sensitive optical coherence tomography,” Opt. Express 21(1), 1163–1180 (2013).

8. M. Yamanari, S. Tsuda, T. Kokubun, Y. Shiga, K. Omodaka, Y. Yokoyama, N. Himori, M. Ryu, S. Kunimatsu-Sanuki, H. Takahashi, K. Maruyama, H. Kunikata, and T. Nakazawa, “Fiber-based polarization-sensitive OCT for birefringence imaging of the anterior eye segment,” Biomed. Opt. Express 6(2), 369–389 (2015).

9. B. Braaf, K. A. Vermeer, M. de Groot, K. V. Vriends, and J. F. de Boer, “Fiber-based polarization-sensitive OCT of the human retina with correction of system polarization distortions,” Biomed. Opt. Express 5(8), 2736–2758 (2014).

10. M. Yamanari, S. Tsuda, T. Kokubun, Y. Shiga, K. Omodaka, N. Aizawa, Y. Yokoyama, N. Himori, S. Kunimatsu-Sanuki, K. Maruyama, H. Kunikata, and T. Nakazawa, “Estimation of Jones matrix, birefringence and entropy using Cloude-Pottier decomposition in polarization-sensitive optical coherence tomography,” Biomed. Opt. Express 7(9), 3551–3573 (2016).

11. S. Jiao and L. V. Wang, “Two-dimensional depth-resolved Mueller matrix of biological tissue measured with double-beam polarization-sensitive optical coherence tomography,” Opt. Lett. 27(2), 101–103 (2002).

12. 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).

13. J. Zhang, W. Jung, J. Nelson, and Z. Chen, “Full range polarization-sensitive Fourier domain optical coherence tomography,” Opt. Express 12(24), 6033–6039 (2004).

14. M. Villiger, K. Otsuka, A. Karanasos, P. Doradla, J. Ren, N. Lippok, M. Shishkov, J. Daemen, R. Diletti, R.-J. van Geuns, F. Zijlstra, G. van Soest, P. Libby, E. Regar, S. K. Nadkarni, and B. E. Bouma, “Coronary plaque microstructure and composition modify optical polarization: A new endogenous contrast mechanism for optical frequency domain imaging,” JACC Cardiovasc. Imaging 11(11), 1666–1676 (2018).

15. 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).

16. 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 using frequency multiplexing,” Opt. Express 16(2), 1096–1103 (2008).

17. 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).

18. 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).

19. 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).

20. L. Duan, T. Marvdashii, and A. K. Ellerbee, “Polarization-sensitive interleaved optical coherence tomography,” Opt. Express 23(10), 13693–13703 (2015).

21. S. Moon and Z. Chen, “Phase-stability optimization of swept-source optical coherence tomography,” Biomed. Opt. Express 9(11), 5280–5295 (2018).

22. W. Choi, B. Potsaid, V. Jayaraman, B. Baumann, I. Grulkowski, J. J. Liu, C. D. Lu, A. E. Cable, D. Huang, J. S. Duker, and J. G. Fujimoto, “Phase-sensitive swept-source optical coherence tomography imaging of the human retina with a vertical cavity surface-emitting laser light source,” Opt. Lett. 38(3), 338–340 (2013).

23. B. Park, M. C. Pierce, B. Cense, S.-H. Yun, M. Mujat, G. Tearney, B. Bouma, and J. de Boer, “Real-time fiber-based multi-functional spectral-domain optical coherence tomography at 1.3 microm,” Opt. Express 13(11), 3931–3944 (2005).
24. B. Vakoc, S. Yun, J. de Boer, G. Tearney, and B. Bouma, “Phase-resolved optical frequency domain imaging,” Opt. Express 13(14), 5483–5493 (2005).
25. B. Baumann, S. O. Baumann, T. Konnegger, M. Pircher, E. Götzinger, F. Schlantz, C. Schütze, H. Sattmann, M. Litschauer, U. Schmidt-Erfurth, and C. K. Hitzenberger, “Polarization sensitive optical coherence tomography of melanin provides intrinsic contrast based on depolarization,” Biomed. Opt. Express 3(7), 1670–1683 (2012).
26. F. Beer, A. Wartak, R. Haindl, M. Gröschl, B. Baumann, M. Pircher, and C. K. Hitzenberger, “Conical scan pattern for enhanced visualization of the human cornea using polarization-sensitive OCT,” Biomed. Opt. Express 8(6), 2906–2923 (2017).
27. B. Baumann, E. Götzinger, M. Pircher, and C. K. Hitzenberger, “Single camera based spectral domain polarization sensitive optical coherence tomography,” Opt. Express 15(5), 2054–2063 (2007).
28. N. J. Smith, S. van der Walt, and E. Firing, “New matplotlib colormaps,” Github repository, (2015).
29. M. Geissbuehler and T. Lasser, “How to display data by color schemes compatible with red-green color perception deficiencies,” Opt. Express 21(8), 9862–9874 (2013).
30. R. Cabanier and N. Andronikos, “Compositing and blending level 1, Candidate recommendation,” W3C (2015). http://www.w3.org/TR/2015/CR-compositing-1-20150113/.
31. E. Collett, Polarized light in fiber optics (SPIE Press, 2003).
32. D. Fried, J. Xie, S. Shafl, J. D. B. Featherstone, T. M. Breunig, and C. Le, “Imaging carries lesions and lesion progression with polarization sensitive optical coherence tomography,” J. Biomed. Opt. 7(4), 618–627 (2002).
33. D. P. Davé and T. E. Milner, “Optical low-coherence reflectometer for differential phase measurement,” Opt. Lett. 25(4), 227–229 (2000).
34. D. P. Davé, T. Akkin, and T. E. Milner, “Polarization-maintaining fiber-based optical low-coherence reflectometer for characterization and ranging of birefringence,” Opt. Lett. 28(19), 1775–1777 (2003).
35. M. K. Al-Qaisi and T. Akkin, “Polarization-sensitive optical coherence tomography based on polarization-maintaining fibers and frequency multiplexing,” Opt. Express 16(17), 13032–13041 (2008).
36. E. Götzinger, B. Baumann, M. Pircher, and C. K. Hitzenberger, “Polarization maintaining fiber based ultra-high resolution spectral domain polarization sensitive optical coherence tomography,” Opt. Express 17(25), 22704–22717 (2009).
37. M. K. Al-Qaisi and T. Akkin, “Swept-source polarization-sensitive optical coherence tomography based on polarization-maintaining fiber,” Opt. Express 18(4), 3392–3403 (2010).
38. M. R. Hee, E. A. Swanson, J. G. Fujimoto and D. Huang, “Polarization-sensitive low-coherence reflectometer for birefringence measurement and ranging,” J. Opt. Soc. Am. B 9(6), 903–908 (1992).
39. M. J. Ju, Y.-J. Hong, S. Makita, Y. Lim, K. Kurokawa, L. Duan, M. Miura, S. Tang, and Y. Yasuno, “Advanced multi-contrast Jones matrix optical coherence tomography for Doppler and polarization sensitive imaging,” Opt. Express 21(16), 19412–19436 (2013).
40. M. Yamanari, S. Makita, Y. Lim, and Y. Yasuno, “Full-range polarization-sensitive swept-source optical coherence tomography by simultaneous transversal and spectral modulation,” Opt. Express 18(13), 13964–13980 (2010).
41. B. Braaf, K. A. Vermees, V. A. D. Sicam, E. van Zeeburg, J. C. van Meurs, and J. F. de Boer, “Phase-stabilized optical frequency domain imaging at 1-μm for the measurement of blood flow in the human choroid,” Opt. Express 19(21), 20886–20903 (2011).
42. R. V. Kuranov, A. B. McElroy, N. Kemp, S. Baranov, J. Taber, M. D. Feldman, and T. E. Milner, “Gas-cell referenced swept source phase sensitive optical coherence tomography,” IEEE Photonics Technol. Lett. 22(20), 1524–1526 (2010).
43. G. Liu, O. Tan, S. S. Gao, A. D. Pechauer, B. Lee, C. D. Lu, J. G. Fujimoto, and D. Huang, “Postprocessing algorithms to minimize fixed-pattern artifact and reduce trigger jitter in swept source optical coherence tomography,” Opt. Express 23(8), 9824–9834 (2015).
44. M. Villiger, D. Lorenser, R. A. McLaughlin, B. C. Quirk, R. W. Kirk, B. E. Bouma, and D. D. Sampson, “Deep tissue volume imaging of birefringence through fibre-optic needle probes for the delineation of breast tumour,” Sci. Rep. 6(1), 28771 (2016).
45. Y.-J. Hong, S. Makita, F. Jaillon, M. J. Ju, E. J. Min, B. H. Lee, M. Itoh, M. Miura, and Y. Yasuno, “High penetration swept source Doppler optical coherence angiography by fully numerical phase stabilization,” Opt. Express 20(3), 2740–2760 (2012).
46. S. Song, J. Xu, S. Men, T. T. Shen, and R. K. Wang, “Robust numerical phase stabilization for long-range swept-source optical coherence tomography,” J. Biophotonics 11(11), 1398–1410 (2017).
47. S. Makita, M. Yamanari, and Y. Yasuno, “Generalized Jones matrix optical coherence tomography: performance and local birefringence imaging,” Opt. Express 18(2), 854–876 (2010).
48. B. I. Akca, B. Považay, A. Alex, K. Wöhrhoff, R. M. de Ridder, W. Drexler, and M. Pollnau, “Miniature spectrometer and beam splitter for an optical coherence tomography on a silicon chip,” Opt. Express 21(14), 16648–16656 (2013).
49. G. Yurtsever, N. Weiss, J. Kalkman, T. G. van Leeuwen, and R. Baets, “Ultra-compact silicon photonic integrated interferometer for swept-source optical coherence tomography,” Opt. Lett. 39(17), 5228–5231 (2014).
50. G. Yurtsever, B. Považay, A. Alex, B. Zabihian, W. Drexler, and R. Baets, “Photonic integrated Mach-Zehnder interferometer with an on-chip reference arm for optical coherence tomography,” Biomed. Opt. Express 5(4), 1050–1061 (2014).
51. S. Schneider, M. Lauermann, P.-I. Dietrich, C. Weimann, W. Freude, and C. Koos, “Optical coherence tomography system mass-producible on a silicon photonic chip,” Opt. Express 24(2), 1573–1586 (2016).

52. Z. Wang, H.-C. Lee, D. Vermeulen, L. Chen, T. Nielsen, S. Y. Park, A. Ghaemi, E. Swanson, C. Doerr, and J. Fujimoto, “Silicon photonic integrated circuit swept-source optical coherence tomography receiver with dual polarization, dual balanced, in-phase and quadrature detection,” Biomed. Opt. Express 6(7), 2562–2574 (2015).

53. I. Grulkowski, M. Gora, M. Szkulmowski, I. Gorczyńska, D. Szlag, S. Marcos, A. Kowalczyk, and M. Wojtkowski, “Anterior segment imaging with Spectral OCT system using a high-speed CMOS camera,” Opt. Express 17(6), 4842–4858 (2009).

54. C. Dai, C. Zhou, S. Fan, Z. Chen, X. Chai, Q. Ren, and S. Jiao, “Optical coherence tomography for whole eye segment imaging,” Opt. Express 20(6), 6109–6115 (2012).

55. M. Ruggeri, S. R. Uhlhorn, C. De Freitas, A. Ho, F. Manns, and J.-M. Parel, “Imaging and full-length biometry of the eye during accommodation using spectral domain OCT with an optical switch,” Biomed. Opt. Express 3(7), 1506–1520 (2012).

56. A. Unterhuber, B. Považay, B. Hermann, H. Sattmann, A. Chavez-Pirson, and W. Drexler, “In vivo retinal optical coherence tomography at 1040 nm - enhanced penetration into the choroid,” Opt. Express 13(9), 3252–3258 (2005).