Effective bidirectional scanning pattern for optical coherence tomography angiography

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Abstract: We demonstrate the utility of a novel scanning method for optical coherence tomography angiography (OCTA). Although raster scanning is commonly used for OCTA imaging, a bidirectional approach would lessen the distortion caused by galvanometer-based scanners as sources continue to increase sweep rates. As shown, a unidirectional raster scan approach has a lower effective scanning time than bidirectional approaches; however, a strictly bidirectional approach causes contrast variation along the B-scan direction due to the non-uniform time interval between B-scans. Therefore, a stepped bidirectional approach is introduced and successfully applied to retinal imaging in normal controls and in a pathological subject with diabetic retinopathy.

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OCIS codes: (170.4460) Ophthalmic optics and devices; (170.4470) Ophthalmology; (170.4500) Optical coherence tomography.

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1. Introduction

Optical coherence tomography angiography (OCTA) is a functional extension of OCT imaging that facilitates the visualization of blood flow in biological tissues. Variants of OCTA methods have been described in recent review articles [1–4]. Through the measurement of temporal variation of the back-scattered OCT signals, the motion of scattering particles in tissue can be detected and used as contrast. In general, OCTA is performed by acquiring a series of OCT data at the same location in the tissue at a constant time interval (Δt). The time interval should be sufficiently long enough to observe changes in the amplitude and/or phase of the OCT signals.

Various lateral scanning devices, such as dual-axis galvanometer-based scanners (GSs) [5–8] and microelectromechanical (MEMS) based resonant scanners [9–13], have been utilized to achieve high speed volumetric OCT imaging. Among them, GSs are the most commonly utilized devices for lateral scanning in OCT imaging techniques because of advantageous features such as precision positioning, controllable scanning frequency, large angle of scan, and a compact device at a reasonable cost. Scanning with GSs commonly proceeds in a raster fashion so that one dimension is scanned rapidly while the other direction is scanned slower. The effectiveness and characteristics of different GS scanning functions (i.e., sawtooth, sinusoidal, and triangular functions) have been theoretically studied and experimentally investigated with an OCT system by Duma et. al. [6, 7]. As demonstrated in their works, the triangular scanning pattern provides images with minimal distortion at a high scanning frequency and large angular amplitudes required for high speed and wide-field OCT imaging. However, these advantages are only valid for bidirectional scans.

By applying the bidirectional scanning method, high speed raster scanning can be performed with a high duty cycle and minimum mechanical stress on the GSs. However, there is a unique characteristic of the bidirectional scan that should be carefully considered for OCTA imaging. Figure 1 shows the time interval of multiple B-scans (a BM-scan) with unidirectional (a) and bidirectional (b) scans driven by sawtooth and triangular functions, respectively. In the case of the unidirectional scan, every depth scan (A-scan) has a constant time interval (Δt) between two successive B-scans. Conversely, in the case of the bidirectional scan the time interval is not constant but changes along the A-scan index. In addition, depending on the B-scan index, the variation of the time interval will either be proportional or inversely proportional to the A-scan index. Conventional OCTA algorithms with two (or more) B-scans per BM-scan generate enhanced vasculature contrast with a constant time interval.

Fig. 1. Comparison between (a) unidirectional and (b) bidirectional scan. In (a), dashed arrow line after each B-scan represents flyback required to restore the scanner to the starting position.
OCTA uses motion contrast to image blood flow without the need for a contrast agent and thereby is extremely motion-sensitive. To minimize artifacts induced by involuntary eye movement, fast data acquisition is critical for OCTA retinal vasculature imaging. In the case of swept source OCT systems, the repetition rate of the source and the sampling density are major elements that determine the image acquisition speed. For OCTA imaging, the number of repeated acquisitions (the number of so called BM-scans) can also improve the imaging speed performance. However, as the data acquisition speed increases, the scanning pattern should be also be considered in order to acquire data efficiently. In this study, we present a swept source based OCTA imaging system using a commercially available 200 kHz swept source. We report on the performance of two representative raster scanning methods, unidirectional (UniDir) and bidirectional (BiDir) scans, and experimentally demonstrated these characteristics for OCT and OCTA imaging. In addition, we propose a new scanning method to effectively acquire OCTA data. To demonstrate its utility, we performed in vivo human retinal imaging on normal and pathological subjects.

2. Methods

2.1 Optical coherence tomography system design

The layout of the OCT system employed in this study is shown in Fig. 2. A commercially available MEMS-based short cavity wavelength swept laser (Axsun Technology Inc. USA) with a center wavelength of 1.064 µm was used as the light source. The light entered the first fiber coupler with a 50:50 coupling ratio; the output ports of the fiber coupler were directed to a phase stabilization interferometer and the OCT interferometer.

Fig. 2. Schematic of 200 kHz swept source OCT and diagram of OCT signals. L, lens; M, mirror; FC, fiber collimator, FR, fiber reflector; PC, polarization controller; BPD, balanced photodetector; DCP, dispersion compensation prism.

The phase stabilization interferometer consisted of a 50:50 fiber coupler and two single mode fiber retroreflectors (PS-1060R-P01-1, Thorlabs Inc. USA). One of the retroreflector ferrules was polished to set the desired optical path difference (OPD) between the two arms of the fiber reflector based interferometer (FRI). Due to the OPD set in the FRI, an additional interference signal (not interfering with the OCT interferometer) was generated at a depth close to DC but not overlapping with the DC term.
The OCT interferometer consisted of a 50:50 fiber coupler. In the reference arm, a dispersion compensation prism was placed to match the dispersion of the sample arm. A custom-built retinal scanner mounted on a slit-lamp base was connected to the sample arm. The retinal scanners consisted of a fiber collimator, a two-axis GS (6210H, Cambridge Technology, USA), and two achromatic doublet lenses. The beam diameter incident on the cornea was calculated to be 1.7 mm (1/e^2). The optical power on the cornea was configured to be less than 1.2 mW, which is lower than the safe exposure limit defined by ANSI at this wavelength [14]. The reflected light from both reference and sample arms were combined at the coupler and directed to a balanced photodetector (PDB460C, Thorlabs Inc. USA). The trigger signal from the swept source laser is first gated by a gate signal that is synchronously generated with the driving signals of the two-axis GSs by a functional generator board (PCIe-6351, National Instruments, Texas). Each gated trigger initiated the analogue-to-digital sampling of a single spectral interference signal corresponding to a single depth-scan. The spectral signal is digitized by a 12-bit waveform digitizer (ATS9350, AlazarTech Inc. Canada) with a 500 MHz sampling frequency and 1024 sampling points. The sampled interference signal was rescaled to the wavenumber domain using a pre-defined rescaling parameter obtained by a time-frequency calibration method [15]. The depth resolution defined by the −3 dB width was measured to be 8.7 µm in air (corresponding to the resolution of 6.3 µm in tissue (n = 1.38)), and the sensitivity measured at ~1 mm depth was 96 dB.

2.2 Scanning profiles and protocols

Figure 3 shows four different raster scanning profiles used in this study. A two BM-scan protocol was employed, and the difference between B-scans within a BM-scan set was used for OCTA calculation. In Fig. 3(a), the constant time interval and the unused portion of the scan due to flyback are observed with the unidirectional scanning profile (UniDir). In Fig. 3(b), the alternating scan direction between sequential B-scans and the absence of flyback are
observed on the bidirectional scanning profile (BiDir). Figure 3(c) shows a new scanning pattern for OCTA with same time intervals between the BM-scan set, named step bidirectional (StepBiDir), in which the slow scan axis is stepped at the B-scan frequency. As a result, the B-scan set within a BM-scan maintains the same scan direction.

The scanning protocols and characteristics of three different scanning methods are provided in Table 1. For all scanning methods, the OCTA volume acquisition time was 1.6-8 s. The flyback portion of the unidirectional scan was experimentally set to 100 A-scans for the given sampling density and scanning range but can be changed as these parameters change. Because of the flyback portion in the B-scan image with the UniDir scanning method, the number of sampling points in an en face image was reduced after cropping the samples acquired during the flyback portion of the scan. The UniDir and StepBiDir scan patterns have a constant time interval between two B-scans in the BM-scan set while the BiDir scan pattern has variable time interval.

| Table 1. Summary of Scanning Protocols and Characteristics |
|------------------------------------------------------------|
| UniDir | BiDir | StepBiDir |
| Number of A/B | 400 | 400 | 400 |
| Number of B/C | 800 | 800 | 800 |
| Flyback | 100 | - | - |
| Δt (ms) | 2 | 0.005-4 | 4 |
| Sampling density | 300x400 | 400x400 | 400x400 |

*Time interval between two B-scans in a BM-scan set; number of points in an en face OCTA image.

2.3 OCTA processing

In this study, a complex differentiation [2] based angiography algorithm was applied for OCTA processing. Since this method requires high stability of the OCT signal phase, a recently developed phase stabilization method (described in Appendix) was used prior to the OCTA processing. The two B-scan OCT signals acquired at the same transverse location can be expressed as

\[
\hat{S}_1(x, z) = A_1(x, z)e^{i\phi_1(x, z)},
\]

\[
\hat{S}_2(x, z) = A_2(x, z)e^{i\phi_2(x, z)},
\]

where \(A_i\) and \(\phi_i\) correspond to the amplitude and phase of the complex OCT signal \(\hat{S}_i\), respectively, of the \(i\)th B-scan. After correcting the axial displacement within the BM-scan set caused by the involuntary movements of the human eye by using a 2D cross-correlation based motion correction algorithm [16], a direct subtraction of two consecutive complex OCT B-scans is calculated as

\[
\Delta A_{flow}(x, z) = |\hat{S}_1(x, z) - \hat{S}_2(x, z)e^{-i\phi_{bulk}(x)}|,
\]

where \(\Delta A_{flow}\) is the amplitude of the differentiated complex OCT signal, which is corresponding to flow signal. Here, \(\phi_{bulk}\) is the bulk phase offset estimated as [17]

\[
\phi_{bulk}(x) = \angle \left( \sum z \hat{S}_1(x, z)\hat{S}_1^*(x, z) \right),
\]

where \(\sum_z\) represents a summation of all pixels along the depth.
2.4 OCTA en face creation

Three-dimensional (3-D) bounded variance smoothing was applied to the motion corrected intensity B-scans in order to reduce the effect of speckle while preserving and enhancing the boundaries between retinal layers. The inner limiting membrane (ILM), posterior surface of the inner plexiform layer (IPL), and posterior boundary of the outer nuclear layer (ONL) were segmented automatically in 3-D using a graph-cut algorithm [18–20]. The angiogram data from the superficial plexiform layer (ILM to IPL), deep plexiform layer (IPL to ONL), and all retinal vascular layers (ILM to ONL) were extracted and averaged in the axial direction to produce projected en face images of the microvasculature. Projection artifacts in the deep layer angiogram were attenuated using a modified slab-subtraction algorithm [14].

3. Results

3.1 Effective scanning duty cycle comparison between scanning methods

The effects of the different scanning methods on OCTA angiograms is shown and discussed in this section. Figure 4 shows a comparison of BM-scans acquired at similar locations using the unidirectional, bidirectional, and step bidirectional scanning techniques. For the unidirectional scans, the uniform scanning direction is clearly seen. In the bidirectional scan, every alternating B-scan is inverted and needs to be flipped before the OCTA could be calculated.

![Fig. 4. Effective duty cycle comparison between (a) UniDir, (b) BiDir, and (c) StepBiDir scanning methods. The yellow solid bars indicate the turnover period, and the solid arrows and dashed lines represent the effective scan and flyback scan, respectively. Scale bar = 400 µm.](image-url)

An additional difference between the scan patterns is the significant contribution of flyback to the total scan time in unidirectional scanning. Because of the flyback portion, the
effective sampling points along the B-scan direction are less than the other scanning methods. For both bidirectional methods, there remains an unusable portion of the image due to the turn-around time of the scanner but the flyback portion is non-existent. Therefore, the number of effective sampling points is higher with both the BiDir and StepBiDir scanning methods. Note that for the BiDir scan method, the time interval between B-scans in a BM-scan is not constant, as illustrated in Fig. 4(b). The time interval between corresponding A-scans of adjacent B-scans changes from 5 µs to 4 ms during the bidirectional scanned data (assuming a 200 kHz swept source).

The StepBiDir scanning method combines the benefits of both the unidirectional and bidirectional methods. As the B-scans within each BM-scan set are acquired in the same direction, the time intervals between corresponding A-scans are constant as shown in Fig. 4. Thus, the uneven contrast distribution of the BiDir method is avoided, and an even contrast distribution comparable to that of the UniDir method is produced. An important distinction is that there is a greater time interval between each corresponding A-scan location in the StepBiDir method than in the UniDir method, (for a given A-scan acquisition rate), because the duration of the backward directed scan was longer than the time required for flyback. It should also be noted that the timing of the step in the slow axis was synchronized with the turn-around point of the fast axis scanning mirror so that there are no additional artifacts aside from the turn-around portion, which is also observed in the BiDir method.

For the images presented in this report, the effective duty cycle of the unidirectional scan, $\eta_U$, was 61.3%, whereas the effective duty cycle of both the bidirectional scans, $\eta_B$, was 84.0% at the given sampling frequency in Table 1. The effective duty cycles were calculated as

$$\eta_U = \frac{W_{\text{eff}}}{W_{\text{total}}} = 100 - \frac{W_{fb} + 2W_{to}}{W_{\text{total}}},$$

(5)

$$\eta_B = \frac{W_{\text{eff}}}{W_{\text{total}}} = 100 - \frac{W_{to}}{W_{\text{total}}},$$

(6)

where $W_{\text{eff}}$ is the effective scanning width (in A-scan pixels), $W_{\text{total}}$ is the total scanning width, $W_{fb}$ is the flyback width and $W_{to}$ is the turnover width caused by switching the scanning direction.

### 3.2 OCTA en face contrast comparison between scanning methods

Figure 5 illustrates the visible contrast differences between the UniDir, BiDir, and StepBiDir scanning methods. OCTA en face images are shown for the summed retinal layers, as well as the superficial and deep plexus layers separately. It should be noted that there is a small amount of random noise present in the images, as averaging across multiple frames was not performed. The duration of the volume acquisition was kept constant across the scanning methods, measured in number of A-scans. Since over 25% of the scan is lost to flyback, the number of sampling points contained in the scanned region is correspondingly less than the equivalent BiDir and StepBiDir scans. The OCTA contrast of the UniDir scan method is consistent along the vertical dimension due to the time interval between each A-scan at a specific location being identical, which is preferable to the non-uniform contrast of the BiDir method. The new StepBiDir method takes advantage of the benefits of both the UniDir and BiDir methods. As the time interval between corresponding A-scans is identical with the proposed method, contrast is uniform throughout the resulting images. Further, the time interval between corresponding A-scans with StepBiDir is longer than that of the UniDir method, resulting in improved contrast that is well illustrated when comparing the slower flow of the deep plexus in Fig. 5(G) and Fig. 5(I). StepBiDir also incorporates the advantage
of the BiDir method as the need for flyback is eliminated and the effective scanning time is decreased. Thus, with this novel scanning protocol we can achieve uniform contrast without the need for flyback and without sacrificing sampling density.

3.3 High quality OCTA imaging performance with StepBiDir

Unique challenges arise when imaging pathological patients as opposed to normal control subjects. For subjects with diabetic retinopathy, motion artifacts due to poor fixation are a common occurrence and contribute to the degradation of the OCTA quality. To help mitigate this issue, we can decrease the time needed for volume acquisition by only using two B-scans per BM-scan, instead of the three B-scans per BM-scan as in our previous work [21–23]. Although this decreases the contrast, by using StepBiDir instead of UniDir the increased time interval between A-scan locations reduces this effect. Additionally, strip-based motion registration and averaging can be used to further enhance contrast and reduce motion artifacts [24]. For the patient imaging protocol, we serially acquired five volumes using the StepBiDir method and processed them through our previously published strip-based registration and averaging algorithm [24]. For the young normal subjects (Fig. 6) there were minimal motion artifacts in the single en face images when compared to the diabetic subject (Fig. 7).
Fig. 6. OCTA contrast comparison of a young control subject between three single StepBiDir en face images, and an average of five serially acquired images. Scale bar = 500 µm.

Fig. 7. OCTA contrast comparison of a patient with diabetic retinopathy between three single StepBiDir en face images, and an average of five serially acquired images. Green circles denote microaneurysms and blue arrows highlight areas of vessel dropout, which are indicators of diabetic retinopathy. Scale bar = 500 µm.
4. Discussion

OCTA blood flow contrast is created by detecting moving particles (e.g. red blood cells) in a tissue volume. To extract blood flow signals for angiogram formation, a time series of OCT measurements at a single location are required. In most current clinical OCTA instruments, these time series OCT measurements are achieved by adopting a multiple B-scan strategy at one location. For increasing signal-to-noise ratio (SNR) of the final angiogram, more repeated B-scans can be used. Consequently, most published high-quality OCTA results were acquired with more than three repeated B-scans at the same transverse location. For example, in the case of a prototype swept source OCTA system (Carl Zeiss Meditec Inc.) [25], B-scans were repeated four times at each position to create high quality angiograms of retinal layers. The AngioVue OCTA system (Optovue, Inc.) utilizes two consecutive orthogonal volume scans with two repeated B-scans, which result in a total of four repeated A-scans at each location [26]. However, this higher image quality comes at the cost of increased image acquisition time and makes the OCTA imaging system more vulnerable to eye motion.

Motion artifacts are a significant issue as they degrade image quality and reduce the informational value of acquired images. In order to minimize involuntary eye motion-related artifacts and/or limitations, effectively faster OCTA data acquisition is required. To improve the acquisition speed, several swept source laser technologies have been developed, which is well summarized by Drexler et al. [27]. As the speed of the swept source is increased, a more efficient scanning strategy is also required. There are many scanning devices that are compatible with high speed swept source OCT; however, as previously mentioned, galvanometer-based scanners still have strong advantages over other implementations. However, applying too high of a frequency to the GS drive function can inflict mechanical damage to the scanner. To minimize this damage, a large portion of the scanning function can be set as flyback, although this results in inefficient OCT and OCTA imaging performances as shown in Fig. 4. The duty cycle of the scan can be maximized by employing a bidirectional scanning protocol, yet, as shown in Fig. 5, the effective OCTA imaging performance is still hindered by the non-constant time delay between corresponding pixels, which also results in contrast variation within a BM-scan.

With the proposed stepped bidirectional scanning method, we can take advantage of the benefits of the unidirectional and bidirectional scan methods. With the stepped bidirectional scanning method, an even OCTA contrast across BM-scan is achieved without the need for flyback. Additionally, the time interval between corresponding A-scans is longer than that of the unidirectional method, which results in an enhanced vascular contrast, particularly at locations of slow flow. Due to the unique characteristics of the stepped bidirectional scanning approach, OCTA imaging could be performed with only two repeated B-scans per BM-scan in order to minimize motion artifacts. In order to increase the contrast, we have registered and averaged the images as shown in Figs. 6 and 7. However, as OCTA contrast depends on both inter-scan time and the number of repeated B-scans, using StepBiDir with two B-scans per BM-scan produces lower contrast than, for example, using a four B-scan per BM-scan conventional bidirectional approach which may need less volumes to achieve the same averaged image quality. There exists a balance between motion artifacts and SNR, and the number of B-scans and inter-scan time are parameters that can be chosen for the given application. The proposed stepped bidirectional approach provides a way to increase the inter-scan time to increase vasculature contrast without increasing the overall volume acquisition time.

Using a 200 kHz A-scan rate swept laser and the StepBiDir scanning method, the time interval between two B-scans in a BM-scan is similar to using a 100 kHz swept laser and the conventional unidirectional raster scanned BM-scan. Another feature of the StepBiDir scanning method is that the time interval between B-scans used for OCTA calculations can be further increased without changing the number of A-scans per B-scan or the number of B-scans per volume. As an example, the time interval can be doubled using the modified
scanning profile shown in Fig. 8. A comparison of the OCTA en face images using the original and modified StepBiDir scanning function is presented in Fig. 9. As speckle decorrelation contrast typically saturates for B-scan intervals longer than ~2ms [28], the time interval between the B-scans is already long enough to clearly visualize the microvasculature with the original StepBiDir scanning profile. As such, there is only a very minor increase in contrast in the capillary areas in the data set acquired with the modified scanning protocol. It should also be noted that increasing the time interval between B-scans may also increase the number of motion artifacts as well as the noise level if the increased time interval is past the decorrelation saturation threshold. However, with higher speed swept sources, there would be a much more noticeable contrast advantage in using the increased time interval StepBiDir protocol without increasing the susceptibility of motion artifacts.

Fig. 8. Time interval increased StepBiDir scanning profile and corresponding B-scan images. Scale bar = 400 µm.

Fig. 9. Images produced by the stepped bidirectional scanning method alternating between two adjacent locations (A-C) and 4 adjacent locations (D-F). By alternating between 4 adjacent locations, the time separation between A-scans was increased. Scale bar = 500 µm.
5. Conclusion

Optical coherence tomography angiography (OCTA) imaging contrast relies on temporal signal changes within the tissue induced by blood flow. Different scanning patterns with which to acquire these OCTA images was investigated in this paper. In the Literature, it has been shown that triangular scanning patterns provide better imaging performance when scanning bidirectionally. However, the time interval is not constant along the A-scan index, which affects the OCTA performance. In this paper, we experimentally demonstrated these bidirectional characteristics with OCTA imaging compared to the conventional unidirectional scan. Through the comparison of results acquired from in vivo human retinal imaging using the different scanning mechanisms, the stepped bidirectional scanning method was found to be best when comparing the en face images of the outer plexiform layer. As the A-scan rate of swept source OCT continues to increase, the stepped bidirectional scanning method in this paper will be able to take advantage of an effective duty cycle (relative to conventional unidirectional raster scanning) while also keeping uniform time intervals between corresponding A-scans.

Appendix: Phase stabilization

In the case of swept source based OCT systems, correcting the phase artifact induced by trigger jitter from the source is essential for utilizing the OCT signal phase information. Unlike amplitude/intensity based OCTA methods, phase or complex signal based OCTA algorithms require additional hardware and/or sophisticated numerical approaches to compensate for the phase artifact. In this study, we used a recently developed phase stabilization algorithm based on previously reported methods [29–33]. The new phase stabilization algorithm requires a simple additional hardware implementation and simple numerical algorithm.

In this phase stabilization process, the relative phase shift between the two spectra to be examined are estimated and compensated. One of the two spectra is denoted as a reference spectrum, and the other as a target spectrum. In the first step for stabilizing the phase, the two digitized spectra are Fourier transformed without the rescaling process and the calibration signals are extracted by a binary window function. Here, the calibration signal is generated from the phase stabilization unit (shown in Fig. 2). The inverse Fourier transformed spectra of the reference and target calibration signals are described as:

\[ S_{cal-ref}(j) = |E_{FR1}(j) + E_{FR2}(j)|^2, \]

\[ S_{cal-tar}(j) = |E_{FR1}(j - \beta) + E_{FR2}(j - \beta)|^2 = S_{cal-ref}(j) \ast \delta(j - \beta), \]

where \( E_{FR1} \) and \( E_{FR2} \) are the sampled spectra of the two calibration beams in the FRI with a sampling index of \( j \). Here, \( \ast \) denotes the convolution operation and \( \beta \) indicates the relative spectral shift amount in the sampling points. Since this sampling index is not along wavenumber \( k \), the \( \beta \) is not linear along the sampling points \( j \). For the second step, the Fourier transformed reference calibration signal is multiplied with the complex conjugate of the Fourier transformed target calibration signal as follows,

\[ \tilde{S}[S_{cal-ref}(j)]\tilde{S}^*[S_{cal-tar}(j)] = \tilde{S}[S_{cal-ref}(j)]\tilde{S}^*[S_{cal-tar}(-j)]\tilde{S}[\delta(-j - \beta)] \]

\[ = \tilde{S}_{cal-tar}(\xi) \tilde{S}_{cal-ref}(\xi) \exp\left[-i2\pi\xi\beta/\tilde{N}\right]. \]
where $\mathcal{I}[]$ represents the discrete Fourier transform, $\xi$ is the index of the discrete Fourier transform of $j$, $N$ is the number of sampling points, the superscript of ‘*’ represents the complex conjugate, and the accent ‘\~’ represents the Fourier transformed signal. Unless two spectra are fully correlated, in which case $S_{\text{cal \_ ref}}(j) = S_{\text{cal \_ tar}}(j)$, the phase difference between two calibration signals can be described as:

$$\Delta \phi(\xi) = \angle \hat{S}_{\text{cal \_ ref}}(\xi) \hat{S}_{\text{cal \_ tar}}^*(\xi) = -i2\pi\xi\beta / N,$$

(10)

As mentioned earlier, the relative phase shift, $\beta$, is not linear along the wavenumber, so the phase difference slope, $\Delta \phi$, cannot be directly used for the spectral shift compensation. Therefore, the phase difference between the two calibration signals at the peak position, $\Delta \phi_p = \Delta \phi(\xi_{\text{peak}})$, is extracted and multiplied to the Fourier transformed OCT signal after wavenumber resampling as follows:

$$\mathcal{I}\left[ S_{\text{OCT \_ ref}}(k) \right] = \mathcal{I}\left[ S_{\text{OCT \_ ref}}(LUT_{j \rightarrow k}(j)) \right] = \hat{S}_{\text{cal \_ ref}}(z)$$

(11)

$$\mathcal{I}\left[ S_{\text{OCT \_ tar}}(k) \right] = \mathcal{I}\left[ S_{\text{OCT \_ tar}}(LUT_{j \rightarrow k}(j)) \right] = \hat{S}_{\text{cal \_ tar}}(z) \exp\left(-i2\pi\Delta \phi_p / z_p z\right)$$

(12)

where

$$LUT_{j \rightarrow k}(j) = \sum_n C_n j^n = C_1 j + C_2 j^2 + C_3 j^3 + \cdots.$$  

(13)

Here, $z_p$ is the depth position of the calibration signal peak, and $LUT_{j \rightarrow k}(j)$ is the polynomial fitting look-up-table for the wavenumber resampling [15]. In the $LUT$, $C$ and $n$ represent the fitting coefficient and order, respectively. Because the phase difference caused by the spectral shift is linear along the wavenumber, the linear phase fitting with the slope ($\Delta \phi_p / z_p$) could be directly applied to the wavenumber rescaled OCT signal.

Similar to Refs [34, 35], the phase stability of our 200 kHz OCTA system with the linear phase fitting was examined by measuring a stationary mirror at different depths without transverse scanning. At each depth, 2,000 a-scans were measured, and the phase noise ($\sigma_{\Delta \phi}$) was calculated by taking the standard deviation (STD) of the phase differences. Figure 10 shows the STD of the phase difference with and without the linear phase fitting as well as the theoretical phase noise described by [34]

$$\sigma_{\Delta \phi} = \sqrt{\left(\frac{1}{\text{SNR}_{\text{OCT}}}\right) + \left(\frac{z_{\text{OCT}}}{z_{\text{cal}}}\right)^2 \left(\frac{1}{\text{SNR}_{\text{cal}}}\right)},$$

(14)

where $\sigma_{\Delta \phi}$ is the STD of the phase difference, $\text{SNR}_{\text{OCT}}$ and $\text{SNR}_{\text{cal}}$ are the SNRs of the OCT and calibration signals, and $z_{\text{OCT}}$ and $z_{\text{cal}}$ are the depth position of the OCT and calibration signals. In this phase stabilization measurement, $\text{SNR}_{\text{OCT}}$ was 45 dB, the roll-off measured at the depth range of 0.18 mm to 2.60 mm was $-3.93$ dB/mm, and $\text{SNR}_{\text{cal}}$ was 41 dB.
The effect of the linear phase fitting based phase stabilization method can be demonstrated by comparing the phase noise without the linear phase fit and the theoretical prediction shown as red box and green line plot in Fig. 10, respectively. For an $\text{SNR}_{\text{OCT}}$ of 45 dB at a depth of ~0.18 mm, $\sigma_{\Delta\phi}$ was measured to be 0.38-degree (6.6-mrad) while its theoretical prediction was 0.34-degree (5.9-mrad). Although compared to the previous report [30], a larger measured phase noise of 5.64-degree (98.4-mrad) was found at the depth of 2.6 mm with $\text{SNR}_{\text{OCT}}$ of 35.5 dB, it is still close to the predicted phase noise of 4.26 degree (94.7-mrad). The main reason for this relatively larger phase noise is that the depth location of the calibration signal is close to the DC. Since this phase stabilization method doesn’t require any cross-correlation calculation, the overall stabilization process could be performed faster than the previous method [30].

**Funding**

Brain Canada; National Sciences and Engineering Research Council of Canada; Canadian Institutes of Health Research; Alzheimer Society Canada; Pacific Alzheimer Research Foundation; Michael Smith Foundation for Health Research; Genome British Columbia.

**Disclosures**

The authors declare that there are no conflicts of interest related to this article.