Buffer-averaging super-continuum source based spectral domain optical coherence tomography for high speed imaging

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Abstract: In super-continuum (SC) source based spectral domain optical coherence tomography (SC-SDOCT), the stability of the power spectral density (PSD) has a significant impact on OCT system sensitivity and image signal to noise ratio (SNR). High speed imaging decreases the camera's exposure time, thus each A-scan contained fewer laser pulse excited SC wideband emissions, resulting in a decrease of SNR. In this manuscript, we present a buffer-averaging SC-SDOCT (BASC-SDOCT) to improve the system's performance without losing imaging speed, taking advantage of the excess output power from typical SC sources. In our proposed technique, the output light from SC was passed through a fiber based light buffering and averaging system to improve the PSD stability by averaging 8 SC emissions. The results showed that 6.96 µs of SC emission after buffering and averaging can achieve the same PSD stability equivalent to a longer exposure time of 55.68 µs, despite increasing the imaging speed from 16.8 kHz to 91.9 kHz. The system sensitivity was improved by 8.6 dB, reaching 100.6 dB, which in turn improved SNR of structural imaging, Doppler OCT velocity measurement, and speckle variance OCT (SVOCT) angiographic imaging as demonstrated by phantom and in vivo experiments.

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1. Introduction

Optical coherence tomography (OCT) [1] technique is an emerging imaging modality for medical diagnostics and treatments. Due to the advantages of non-invasiveness, high resolution and high imaging speed, OCT is widely used for disease diagnosis on various tissues, such as retina, cardiology and dermatology. In addition to microstructural imaging, OCT microvasculature images are also widely used in medical imaging and play an increasingly important role with algorithms for extracting blood flow information such as optical Doppler tomography (ODT) [2,3] or color Doppler OCT (CDOCT) [4] which are able to calculate the axial velocity component of moving scattering particles. Morphological OCT microvasculature imaging, collectively termed OCT angiography (OCTA), also performs a key role for disease diagnosis in clinical applications. A number of OCTA algorithms have been presented, such as speckle variance OCT (SVOCT) [5–8], split-spectrum amplitude-decorrelation angiography (SSADA) [9], and ultrahigh sensitivity optical microangiography (UHS-OMAG) [10–12] - A common theme in all of these techniques is the requirement of either high A-scan or B-scan density for repetitive imaging of target tissue to ascertain difference between stationary versus moving particles, requiring high speed imaging.
Recently, super-continuum (SC) source has been used in OCT field more widely due to large bandwidth (covering from 350 to 2350 nm) and high output power (>1W). As axial resolution in OCT is determined by bandwidth and center wavelength, SC offers an important option for super-high resolution OCT [13–17], such as the axial resolution of 0.88 µm achieved in Lichtenegger's recent work [14]. High bandwidth also makes it possible to have a dual-band OCT for tissue imaging at the same time [18], which provides comparison of image capabilities between different spectral bands. Meanwhile, high light power enables high speed and high sensitivity parallel OCT imaging, where an A-scan rate of 1,024k lines/s was achieved in Barrick's work [19]. For microvasculature imaging, SC source was successfully used to reconstruct high-resolution micro-angiography of mouse ear [20], where multiple A-scans were required for each position imaged.

In SC light sources, high bandwidth and high output power are achieved by optical nonlinear processes acted upon pump pulses from a seed laser [21], therefore SC's power spectral density (PSD) is less stable compared to traditional diode laser [22]. Compared to superluminescent diode (SLD) or mode-locked swept source, SC-SDOCT suffers the trade-off between exposure time (A-scan rate) and system sensitivity: longer exposure time can improve PSD stability, because light excited by more pump pulses is accumulated on camera for each exposure [23]. Of course, longer exposure time would lead to slower frame rate. Finally, system sensitivity and structural OCT image signal-to-noise ratio (SNR) contribute to phase noise in Doppler OCT or blood flow measurements.

In the published literature, there had been incomplete reporting of seed laser pulse repetition rate for SC light sources with measured sensitivity and camera exposure time. In addition, when feedback based technique is used to improve the power stability of commercial SC sources (e.g., SuperK Extreme, NKT Photonics), the applicable wavelength has been in the visible to ~900 nm range, which leaves the 1300 nm range sub-optimally stabilized, where many OCT applications reside. Nevertheless, the typical sensitivity range had been 87 to 98 dB at about 20 µs exposure time without clear reporting of seed laser frequency [18,20,24–26]. An exception has been Yuan et al [16], who reported system sensitivity of 108 dB for structural imaging at exposure time of approximately 14.3 µs and center wavelength of 900 nm. This was achieved using high seed laser frequency of 325 MHz, which meant there were 4,647 pulses per camera exposure.

In this study, we investigated and analyzed the unstable PSD noise of the suboptimal infrared spectral region of a commercial SC source (SuperK Extreme, NKT Photonics). At a comparatively low seed frequency of 80 MHz, a sensitivity of 100.6 dB was achieved with an A-line rate of 91.9 kHz at camera exposure time of 6.96 µs (approximately 4,454 seed laser pulses per exposure). To our knowledge, this is the fastest A-scan rate and shortest exposure time achieved with a comparable sensitivity, using a low-cost, passively buffered-averaging SC-SDOCT (BASC-SDOCT) technique to improve system performance and maintained high imaging speed.

2. Methods

2.1 Fiber based light buffering and averaging system

In SC-SDOCT, camera accumulated the light excited by the pump pulses from seed laser within the exposure time period, so by increasing exposure time (slower A-scan speed), SC’s PSD stability can be improved because more pulses can be detected. This trade-off can be circumvented by using a fiber based light buffering and averaging system, as shown in Fig. 1. Output light from SC was first split by a 1 × 2 coupler, half of which was further split into four and then buffered by four segments of fiber with length of 0, \(L_0\), 2\(L_0\) and 3\(L_0\), the delayed beams were then combined by a 1 × 4 coupler. The other half of the output light from the initial 1 × 2 coupler was buffered by a segment of fiber with length of 4\(L_0\) then split into four as well with similar delay fibers and combined. A final 1 × 2 coupler was used to combine the two beams from the last two 1 × 4 couplers. To allow averaging, the differential delay time
caused by the fiber length \((L_0)\) has to be equal to or longer than the camera exposure time. Therefore, the minimum fiber length \((L_0)\) can be calculated by

\[ L_0 = \frac{c \cdot T}{n}, \]

where \(c\) is speed of light, \(T\) is camera's exposure time and \(n\) is fiber's refractive index.

![Schematic of fiber based light buffering and averaging system.](image)

Fig. 1. Schematic of fiber based light buffering and averaging system.

The above passive fiber buffering and averaging network effectively allows 8 times more seed laser pulses to enter the OCT interferometer while maintaining the same camera exposure time for high speed imaging. The main disadvantage is power loss through the network, including 9 dB of mainly coupler, connector, and fiber propagation losses, although typical SC light sources have significantly higher output power than necessary for biological imaging.

2.2 Simulations

We first simulated the PSD averaging process to investigate the performance of the proposed technique. In SC-SDOCT, except for the noises discussed in Ref [27], unstable PSD could also introduce noise which could be marked as \(\sigma^2_{sc}\), and the total noise after Fourier transform can then be express by

\[ \bar{\sigma}^2 = \frac{1}{N}(\sigma^2_{shot} + \sigma^2_{excess} + \sigma^2_{dark} + \sigma^2_{read} + \sigma^2_{sc}), \]

where \(N\) is the pixel number of camera. For the next four terms, \(\sigma^2_{shot}\) is shot noise, \(\sigma^2_{excess}\) is excess noise, \(\sigma^2_{dark}\) and \(\sigma^2_{read}\) are dark noise and readout noise of the camera. The first four are held constant when the provided input optical power for SDOCT is the same. We assume the SC noise as a Gaussian distribution with standard deviation (STD) of \(\sigma^2_{sc}\) shown in Fig. 2(b) according to the measured original PSD in Fig. 5 and the measured noise intensity in Fig. 6. Figure 2(c) and Fig. 2(d) showed the overlapped 1000 A-scans of PSD without and with buffer-averaging, respectively. The linear interference fringes to pixel index number for a mirror in sample arm could then be obtained through multiplying PSD by a cosine function, as shown in Fig. 2(e) and Fig. 2(f). After performing FFT on the fringes, the averaged point spread function (PSF) in log-scale of intensity for both with and without buffer-averaging.
were shown in Fig. 2(g). By performing the phase sensitive Doppler algorithm on adjacent two A-scans, Doppler phase shift images could be obtained and the histograms of Doppler phase shifts at the pixels with maximum intensity were shown in Fig. 2(h) and Fig. 2(i). It was found that not only PSF noise floor was decreased, the STD of Doppler phase shifts was also improved by 2.6 times using 8 times buffer-averaged PSD, which matches the theoretically predicted improvement of 2.8 times.

![Simulation results](image)

**Fig. 2. Simulation results.** (a) Theoretical PSD. (b) Plot of noise STD versus optical intensity. (c)-(d) Overlapped 1000 A-scans of original PSD and 8-buffer–averaged PSD. (e)-(f) Overlapped 1000 A-scans of simulated interference fringes obtained by multiplying a cosine function by (c)-(d). (g) Averaged PSF of 1000 PSFs obtained by performing FFT on (e) and (f). (h)-(i) Histograms of Doppler phase shifts (at peak intensity pixel) between each two adjacent A-scans. In (g)-(i), blue data and red data are from the original PSD and buffer-averaged PSD, respectively.

3. Experimental set-up of BASC-SDOCT and results

3.1 BASC-SDOCT set-up and scanning protocol

The schematic of BASC-SDOCT was shown in Fig. 3, where SC was a commercial laser (SuperK Extreme EXR-4, NKT Photonics) which has a repetition of 78 MHz and a wavelength range spanning 400-2400 nm. The infrared portion of SC with wavelength range of 1100-1500 nm was obtained by SuperK Gauss box (NKT Photonics) and then forwarded to fiber based light buffering and averaging system. After that, the buffer-averaged beam went into SDOCT through a circulator. The structure of interferometer was a 2 × 2 coupler with split ratio of 90:10, 90% of light power was forwarded to sample arm while the other was forwarded to reference arm. The reflected light from reference arm and backscattered light from sample were combined by the coupler and forwarded to spectrometer. Lens 1 and lens 2
are both collimators (F280APC-C, Thorlabs), and lens 3 has a focus length of 30 mm, giving a lateral resolution of 15 µm. The NIR spectrometer (P&P Optica Inc., Waterloo, Canada) based on a grating with frequency of 892 lines/mm has a spectral resolution of 0.365 nm and a wavelength range of 1190-1370 nm, achieving an axial resolution of 8 µm. The camera (Goodrich SU-LDH2) has 1024 pixels and 12 bit depth, with maximum A-scan rate at 91.9 kHz, exposure time of 6.96 µs, and data readout time for each A-scan at 3.94 µs. Single mode fiber (SMF-28) is used in our buffer-averaging network and the fiber length $L_0$ is 1.5 km.

In this work we tested the performance of the proposed technique on structural images, Doppler phase shift images and angiographic images of SVOCT. For structural images and SVOCT images, the scanning protocol is regular raster scanning pattern, where 1000 A-scans were acquired within each B-scan and duty cycle of 83.1% was used for fast scanning. However, the minimum measureable velocity of CDOCT technique with A-scan rate of 91.9 kHz was too high for blood flow of human skin capillary. We performed a zigzag scanning protocol for CDOCT instead of decreasing A-scan rate. Figure 4(a) showed the scanning pattern, where B-scan range was evenly portioned into three parts, each part was scanned 10 times. The images from 3 different parts were finally stitched into one B-scan image of large scanning range. We scanned a tilted mirror to test the performance of the zigzag scanning protocol as shown in Fig. 4(d).

For structural imaging and SVOCT imaging (Fig. 6, Fig. 8, and Fig. 11), the scanning protocol is regular raster scanning with duty cycle of 83.1%, where 1000 A-scans are acquired within each B-scan. Each position was scanned more than once to calculate inter-frame based SVOCT images [6]. For Doppler velocity calculation (Fig. 7, Fig. 9 and Fig. 10), zigzag scanning protocol was performed in the first two settings, where 100 A-scans were acquired within each small portion of B-scan and Doppler phase shift was calculated between each two adjacent frames of 10 frames. Then, Kasai algorithm [3] with a window of 9 (phase shift frames) × 5 (depth direction) pixels was performed on the obtained 9 frames of phase shift images to suppress the phase noise. In the last setting with exposure time of 55.68 µs, the scanning pattern was also the regular raster scanning pattern as the same with structural imaging, and phase shift was calculated between two adjacent A-scans. The reason of not using zigzag scanning pattern is that severe bulk motion noise will occur during the scanning of in vivo experiments due to small A-scan rate. The time intervals between two frames in
raster scanning and between two small portion frames in zigzag scanning are 13 ms and 1.3 ms, respectively.

Fig. 4. Schematic of scanning protocol for CDOCT technique. (a) Synchronized trigger signal for camera and driving signal of fast scanning galvo. (b) Structure of tilted mirror scanning. (c) Image without stitching. (d) Stitched images.

3.2 Performance of BASC-SDOCT

The performance of BASC-SDOCT was tested by inserting a mirror in sample arm, where SDOCT worked in M-mode (both x- and y-galvos were kept still). Because PSD averaging and increasing exposure time could both increase PSD stability, we also measured the results with exposure time of 55.68 µs (8 times longer) for comparison in all experiments of this work. The results were shown in Fig. 5. By dividing the STD of intensity by the mean intensity at each pixel, the relative noise plots could be obtained and shown Fig. 5(d). It was found that the noise intensities of 8-buffer-averaged PSD and the PSD with exposure time of 55.68 µs were comparable, and approximately 2.5 times of improvement as calculated by dividing the blue by the green curves was achieved by buffer-averaging method. Figure 5(e)-(g) were the relative PSFs (log-scale amplitude) with the intensity divided by noise floor (STD of background). Figure 5(h) showed the comparison of the plots of maximum value of relative PSFs versus depths, where 8-buffer-averaged PSD and the PSD with exposure time of 55.68 µs achieved higher performance. For OCT angiographic techniques, such as SVOCT, repeatability was one of the most important factors because intensities of different A-scans were used for extracting dynamic signals from background, and a low repeatability would increase background noise level and result in a decrease of sensitivity. Here, repeatability was evaluated by calculating the histograms of PSF's differential peak values between two adjacent PSFs and the results were shown in Fig. 5(l), where repeatability was improved about 2 times by using 8-buffer-averaged PSD and the PSD with exposure time of 55.68 µs. CDOCT was performed on two adjacent A-scans to calculate phase shifts. The obtained plots of phase shift STDs versus SNR were calculated and shown in Fig. 5(p). An adjustable attenuator was inserted in sample arm to modulate reflected light intensity, achieving SNR variation, and the SNR of PSF was calculated by
\[ \text{SNR} = 20 \cdot \log_{10} \left( \frac{I_{\text{max}}}{\sigma_{\text{bg}}} \right), \]

(3)

where \( I_{\text{max}} \) is the peak value of PSF and \( \sigma_{\text{bg}} \) is STD of background. It was found that phase shift STDs were reduced of 1.4 times (quotient of STDs of two scenarios at the same SNR) compared to original PSD.

| Exposure time: 6.96 \( \mu \)s | Exposure time: 6.96 \( \mu \)s with 8 times of averaging | Exposure time: 55.68 \( \mu \)s | Comparison |
|---------------------------------|-------------------------------------------------|-------------------------------|-----------|
| Spectral plot                   | Spectral plot                                   | Spectral plot                 | Spectral plot |
| ![Spectral plot](image1)        | ![Spectral plot](image2)                        | ![Spectral plot](image3)     | ![Spectral plot](image4) |
| Intensity vs. Wavelength [nm]   | Intensity vs. Wavelength [nm]                   | Intensity vs. Wavelength [nm] | Intensity vs. Wavelength [nm] |
|                                 |                                                 |                               | Comparison |
| Relative noise plots            | Relative noise plots                             | Relative noise plots          | Relative noise plots |
| ![Relative noise plots](image5) | ![Relative noise plots](image6)                 | ![Relative noise plots](image7) | ![Relative noise plots](image8) |
| Mean of STD/Intensity [sa]      | Mean of STD/Intensity [sa]                      | Mean of STD/Intensity [sa]    | Mean of STD/Intensity [sa] |
| Depth [\( \mu \)m]             | Depth [\( \mu \)m]                              | Depth [\( \mu \)m]           | Depth [\( \mu \)m] |
| Pixel count                     | Pixel count                                     | Pixel count                   | Pixel count |
| ![Pixel count](image9)          | ![Pixel count](image10)                         | ![Pixel count](image11)      | ![Pixel count](image12) |
| Differential intensity          | Differential intensity                           | Differential intensity        | Differential intensity |
| Phase shift plots               | Phase shift plots                               | Phase shift plots             | Phase shift plots |
| ![Phase shift plots](image13)   | ![Phase shift plots](image14)                   | ![Phase shift plots](image15) | ![Phase shift plots](image16) |
| Pixel count                     | Pixel count                                     | Pixel count                   | Pixel count |
| Phase shift [rad]               | Phase shift [rad]                               | Phase shift [rad]             | Phase shift [rad] |
| ![Phase shift plots](image17)   | ![Phase shift plots](image18)                   | ![Phase shift plots](image19) | ![Phase shift plots](image20) |
| SNR [dB]                        | SNR [dB]                                        | SNR [dB]                      | SNR [dB] |

Fig. 5. Performance of BASC-SDOCT. (a)-(c) Overlapped 1000 A-scans of PSD under three different settings. (d) Relative noise plots obtained through dividing STD by the mean value at each pixel. (e)-(g) Plots of relative PSFs with the intensity at each pixel divided by noise floor (STD of background). (h) Plots of SNR versus depth. (i)-(k) Histograms of differential peak values of 1000 PSFs at the depth marked by a black arrow in (l). (l) Plots of the STDs of PSFs’ differential peak values versus depth. (m)-(o) Histograms of the phase shifts of PSF peak value pixel at the position marked by a black arrow in (p). (p) The plots of phase-shift STDs versus SNRs. In (d), (h), (l) and (p), blue, green and red data are from the original PSD, 8-buffer-averaged PSD and the PSD with 8 times longer exposure time.

To analyze the noise reduced by our proposed method, the measured camera dark noise and total noise were shown in Fig. 6. Dark noise was measured by calculating the STD at each pixel of 1000 A-scans acquired with the laser turned off. By blocking sample arm, 1000 A-scans of original PSD, 8-buffer-averaged PSD and the PSD from a SLD were acquired separately, for which the plots of STD versus mean intensity at each pixel were shown in Fig. 6(b). Experimentally, SLD generated the lowest noise floor which included characteristics of
the camera CCD and interferometer, such as dark noise, excess noise, readout noise and shot noise. The SC noise ($\sigma_{sc}$) could be estimated by comparing to the SLD noise floor as a benchmark. It was found that SC noise dominated the total noise and most of SC noise was reduced after 8 times of buffer-averaging.

We further measured the sensitivity of SDOCT with and without using the proposed method by inserting an optical neutral filter with OD = 3 (60 dB) in sample arm. The measured averaged PSFs were shown in Fig. 7, where the sensitivity could be improved to 100.6 dB which was approximately 8.6 dB higher than original PSD at the majority of imaging depths.

3.3 Phantom results

We scanned an orange to verify the performance of BASC-SDOCT on structural imaging and the results were shown in Fig. 8. Scanning protocol was regular raster scanning and 1000 A-scans were acquired within 2 mm. In comparison of Fig. 8(a)-(f), it was found that the 8-buffer-averaged PSD and PSD of 8 times longer exposure time could both detect more structural information, such as those marked by yellow arrows. In Fig. 8(g), intensity plots at the positions marked by dashed green lines in Fig. 8(d)-(f) were shown and SNRs were calculated for quantitative comparison.
We then scanned a flow phantom to test the performance of BASC-SDOCT for CDOCT and SVOCT techniques. The phantom was made of silicone gel with TiO$_2$ and a plastic tube with internal diameter of 0.28 mm was buried in the silicone gel to simulate vessels with Doppler angle of 80°. Intralipid solution at 5% was pumped by a syringe pump (Harvard Apparatus, Holliston, MA) to flow through the tube. For CDOCT, the scanning protocol of Fig. 4 was performed and 3000 A-scans were acquired for each full B-scan. The average flow rate of intralipid solution was 0.2 ml/h with mean velocity of 0.9 mm/s for original PSD and
8-buffer-averaged PSD with time interval of 1.3 ms, and 3 ml/h for the PSD with exposure time of 55.68 µs, where Doppler phase shift was calculated between two adjacent A-scans. For SVOCT, scanning protocol was regular raster scanning pattern, two B-scans at the same position were acquired for calculating speckle variance. For each B-scan, 1000 A-scans were acquired and the syringe pump was stopped. CDOCT images and SVOCT images were shown in Fig. 9 and Fig. 10, respectively.

Fig. 9. CDOCT images of flow phantom. (a) Structural image. (b)-(d) Doppler phase shift images of original PSD of 6.96 µs, 8-buffer-averaged PSD and the PSD with exposure time of 55.68 µs, where a Kasai window of 5 × 9 pixels were used [3]. (e) Phase shift plots at the positions marked by dashed red lines in (b)-(d), where blue, red and green curves are data of original PSD of 6.96 µs, 8-buffer-averaged PSD and the PSD with exposure time of 55.68 µs. The regions marked by dashed white rectangles were used as background for SNR calculation. (b)-(d) share the same scale bar.

In Fig. 9(e), phase shift plots at the positions marked by dashed red lines in Fig. 9(b)-(d) were shown and SNR were calculated by using Eq. (3) for quantitative comparison. The results demonstrated that the 8-buffer-averaged PSD and the PSD with exposure time of 55.68 µs achieved comparable SNRs and both were higher than original PSD. Figure 10(e) showed intensity plots of SVOCT images, where curves were normalized to mean value of intensities within the black dashed rectangles, the mean value was also used as the signal intensity for SNR calculation in Eq. (3). The zoomed image in Fig. 10(e) showed that 8-buffer-averaged PSD achieved better background noise level and SNR than original PSD.
3.4 Biological tissue results

We scanned both *ex vivo* chicken thigh muscle and *in vivo* human skin from a female volunteer's finger to further verify the performance of BASC-SDOCT system, the scanning protocol of Fig. 4 was performed for original PSD of 6.96 µs and 8-buffer-averaged PSD, and 3000 A-scans were acquired for each full B-scan. For the PSD with exposure time of 55.68 µs, regular raster scanning was performed and the Doppler phase shift was calculated between adjacent A-lines. The results of *ex vivo* chicken thigh muscle and *in vivo* skin are shown in Fig. 11 and Fig. 12, respectively.
In comparison of Fig. 11(b)-(d) and Fig. 12(b)-(d), it was found that deeper structural information was better reconstructed by the buffer-averaged PSD and the PSD with exposure time of 55.68 µs due to reduced background noise floor, such as the regions marked by yellow arrows. Figure 11(e)-(g) and Fig. 12(e)-(g) were the calculated Doppler phase shift images at the same positions with the structural images, where the signals marked by white arrows in Fig. 12(e)-(g) are blood flow signals. To quantitatively compare the performance of CDOCT technique, the histograms and STDs of the background phase shifts at the regions marked by dashed white rectangles and lines were shown and calculated. The results demonstrated that buffer-averaged PSD and the PSD with exposure time of 55.68 µs achieved comparable sensitivity and both were higher than original PSD of 6.96 µs.
Fig. 12. Cross sectional human skin images. (a) Photograph of a volunteer’s left hand and the region marked by a red line was scanned. (b)-(d) Structural images of original PSD, 8-buffer-averaged PSD and the PSD with exposure time of 55.68 µs. (e)-(g) Doppler phase shift images of (b)-(d), respectively, where a Kasai window of 5 × 9 pixels was used. (h)-(j) Histograms of the phase shifts at the positions marked by dashed white lines (indicative of the background level) in (e)-(g), respectively. (b)-(d) and (e)-(g) share the same scale bar.

We also scanned a local region of 2 × 2 mm² on the volunteer's finger to evaluate the performance of BASC-SDOCT for SVOCT, where scanning protocols for the three settings were all regular raster scanning, where 1000 A-scans were acquired for each B-scan and each position was scanned 4 times. During post data processing of all the three data sets, three cross sectional SVOCT images were averaged to improve image quality, and maximum intensity projection was performed to calculate en face SVOCT images. The results were shown in Fig. 13 and demonstrated that 8-buffer-averaged PSD achieved the best image quality. Compared to the PSD with exposure time of 55.68 µs, higher image contrast was achieved by faster scanning speed, which suppressed bulk motion noise effectively.
Fig. 13. En face SVOCT images within a local region of $2 \times 2$ mm$^2$. (a)-(c) En face SVOCT images of original PSD, 8-buffer-averaged PSD and the PSD with exposure time of 55.68 $\mu$s within depth range of 390-900 $\mu$m. (d)-(f) Zoomed images within the local region marked by dashed white rectangles in (a)-(c), respectively. (g) Plots of SVOCT signals at the positions marked by white dashed lines in (d)-(f), where blue, red and green curves are data of original PSD, 8-buffer-averaged PSD and the PSD with exposure time of 55.68 $\mu$s. (a)-(c) share the same scale bar.

4. Discussions and conclusion

In this work, output power of infrared portion from SC was 190 mW, and 10 mw (approximately 7.7 mw within the illumination band of our OCT) was obtained for 8-buffer-averaged PSD. The output power is expected to be higher if fiber was spliced rather than connected using FC/APC connectors. To avoid saturating the CCD camera and maintaining similar readout counts under different experimental conditions, the optical powers forwarded into the SDOCT were attenuated to 10 mw for original PSD with exposure time of 6.96 $\mu$s and 1.25 mw for the PSD with exposure time of 55.68 $\mu$s by attenuators, respectively.

One limitation of the proposed buffer averaging technique is that light power is reduced more than 9 dB, so the proposed passive technique is not suitable for the applications where high power is needed, such as parallel imaging [19]. However, the proposed technique provided an excellent chance to improve SC’s performance, utilizing otherwise attenuated excessive high power for regular spot scanning OCT, because optical power of $<$20 mw is usually used for skin imaging and microwatt level is usually used for retinal imaging. Active amplification of the light after buffer-averaging could be explored in the future, although additional noise terms would be introduced.

The key advantage of the proposed technique is that the performance of SC-SDOCT could be improved without losing A-scan speed which is a key factor for in vivo angiographic imaging. Although the same performance could be achieved by increasing exposure time, heavy bulk motion noise will be caused by slow scanning speed, resulting in a decrease of reliability for disease diagnosis in clinical OCTA application. Furthermore, in commercial SC source, such as NKT photonic, feedback based power stable process is performed on visible
to ~900 nm wavelength range, which also poses a challenge to use infrared portion for high speed and high quality imaging. The proposed technique provided a simple solution for this problem, where at a comparatively low seed frequency of 80 MHz, we achieved a sensitivity of 100.6 dB with an A-line rate of 91.9 kHz at camera exposure time of 6.96 µs. To our knowledge, this is the fastest A-scan rate and shortest exposure time achieved with a comparable sensitivity, using a low-cost, passively buffered-averaging SC-SDOCT (BASC-SDOCT) technique to passively improve system performance and maintained high imaging speed.

Temporal de-correlation of the light source contributes to the background noise in CDOCT and SVOCT. For inter-frame based CDOCT and SVOCT adopted in this work, the proposed technique could achieve better sensitivity due to the improved PSD stability. And by using our technique for inter-A-line based CDOCT and SVOCT, input light for two consecutive A-scans is highly correlated because 7/8 of the seed laser pulses are present in both A-scans. Any blood flow induced perturbations in the sample arm, causing Doppler shift or speckle variance, would modulate the detected interference fringes. The improved correlation from the input light after buffer-averaging is one of the contributing factors for decreased noise floors in both CDOCT and SVOCT in comparison to original SC source without buffering.

In conclusion, we proposed a buffer-averaging super-continuum source based spectral domain optical coherence tomography (BASC-SDOCT) for high speed imaging, where output light from SC was buffered and averaged to improve spectral stability, resulting in an improvement of system performance. The proposed technique is cheaper or faster compared to increasing seed laser's repetition rate or using longer exposure time. Phantom experiments and in vivo experiments were both implemented to verify the performance of BASC-SDOCT, and all results demonstrated that the proposed technique achieved higher sensitivity for structural images and higher SNR for CDOCT and SVOCT images without losing imaging speed.

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References
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. Puliafito, and J. G. Fujimoto, “Optical coherence tomography,” Science 254(5035), 1178–1181 (1991).
2. Z. Chen, T. E. Milner, S. Srinivas, X. Wang, A. Malekafzali, M. J. C. van Gemert, and J. S. Nelson, “Noninvasive imaging of in vivo blood flow velocity using optical Doppler tomography,” Opt. Lett. 22(14), 1119–1121 (1997).
3. V. Yang, M. Gordon, B. Qi, J. Pekar, S. Lo, E. Seng-Yue, A. Mok, B. Wilson, and I. Vitkin, “High speed, wide velocity dynamic range Doppler optical coherence tomography (Part I): System design, signal processing, and performance,” Opt. Express 11(7), 794–809 (2003).
4. J. A. Izatt, M. D. Kulkarni, S. Yazdanfar, J. K. Barton, and A. J. Welch, “In vivo bidirectional color Doppler flow imaging of picoliter blood volumes using optical coherence tomography,” Opt. Lett. 22(18), 1439–1441 (1997).
5. J. Barton and S. Stromski, “Flow measurement without phase information in optical coherence tomography images,” Opt. Express 13(14), 5234–5239 (2005).
6. A. Mariampillai, B. A. Standish, E. H. Moriyma, M. Khurana, N. R. Munce, M. K. K. Leung, J. Jiang, A. Cable, B. C. Wilson, I. A. Vitkin, and V. X. D. Yang, “Speckle variance detection of microvasculature using swept-source optical coherence tomography,” Opt. Lett. 33(13), 1530–1532 (2008).
7. A. Mariampillai, M. K. K. Leung, M. Jarvi, B. A. Standish, K. Lee, B. C. Wilson, A. Vitkin, and V. X. D. Yang, “Optimized speckle variance OCT imaging of microvasculature,” Opt. Lett. 35(8), 1257–1259 (2010).
8. C. Chen, K. H. Y. Cheng, R. Jakubovic, J. Jivraj, J. Ramjist, R. Deorajh, W. Gao, E. Barnes, L. Chin, and V. X. D. Yang, “High speed, wide velocity dynamic range Doppler optical coherence tomography (Part V): Optimal
utilization of multi-beam scanning for Doppler and speckle variance microvascular imaging,” Opt. Express 25(7), 7761–7777 (2017).
9. Y. Jia, O. Tan, J. Tokayer, B. Potsaid, Y. Wang, J. J. Liu, M. F. Kraus, H. Subhash, J. G. Fujimoto, J. Hornegger, and D. Huang, “Split-spectrum amplitude-decorrelation angiography with optical coherence tomography,” Opt. Express 20(4), 4710–4725 (2012).
10. L. An, J. Qin, and R. K. Wang, “Ultrahigh sensitive optical microangiography for in vivo imaging of microcirculations within human skin tissue beds,” Opt. Express 18(8), 8220–8228 (2010).
11. L. An, T. T. Shen, and R. K. Wang, “Using ultrahigh sensitive optical microangiography to achieve comprehensive depth resolved microvasculature mapping for human retina,” J. Biomed. Opt. 16(10), 106013 (2011).
12. S. Yousefi, J. Qin, and R. K. Wang, “Super-resolution spectral estimation of optical micro-angiography for quantifying blood flow within microcirculatory tissue beds in vivo,” Biomed. Opt. Express 4(7), 1214–1228 (2013).
13. S. P. Chong, M. Bernucci, H. Radhakrishnan, and V. J. Srinivasan, “Structural and functional human retinal imaging with a fiber-based visible light OCT ophthalmoscope,” Biomed. Opt. Express 8(1), 323–337 (2017).
14. A. Lichtenegger, D. J. Harper, M. Augustin, P. Eugui, M. Muck, J. Gesperger, C. K. Hitzenberger, A. Woehrler, and B. Baumann, “Spectroscopic imaging with spectral domain visible light optical coherence microscopy in Alzheimer’s disease brain samples,” Biomed. Opt. Express 8(9), 4007–4025 (2017).
15. M. Maria, I. Bravo Gonzalo, T. Feuchter, M. Denninger, P. M. Moselund, L. Leick, O. Bang, and A. Podoleanu, “Q-switch-pumped supercontinuum for ultra-high resolution optical coherence tomography,” Opt. Lett. 42(22), 4744–4747 (2017).
16. W. Yuan, J. Mavadia-Shukla, J. Xi, W. Liang, X. Yu, S. Yu, and X. Li, “Optimal operational conditions for supercontinuum-based ultrahigh-resolution endoscopic OCT imaging,” Opt. Lett. 41(2), 250–253 (2016).
17. K. Q. Kieu, J. Klein, A. Evans, J. K. Barton, and N. Peyghambarian, “Ultrahigh resolution all-reflective optical coherence tomography system with a compact fiber-based supercontinuum source,” J. Biomed. Opt. 16(10), 106004 (2011).
18. S. Chen, X. Shu, J. Yi, A. Fawzi, and H. F. Zhang, “Dual-band optical coherence tomography using a single supercontinuum laser source,” J. Biomed. Opt. 21(6), 066013 (2016).
19. J. Barrick, A. Doblas, M. R. Gardner, P. R. Sears, L. E. Ostrowski, and A. L. Oldenburg, “High-speed and high-sensitivity parallel spectral-domain optical coherence tomography using a supercontinuum light source,” Opt. Lett. 41(24), 5620–5623 (2016).
20. Z. Zhi, J. Qin, L. An, and R. K. Wang, “Supercontinuum light source enables in vivo optical microangiography of capillary vessels within tissue beds,” Opt. Lett. 36(16), 3169–3171 (2011).
21. J. M. Dudley and J. R. Taylor, Supercontinuum Generation in Optical Fibers (Cambridge University Press, 2010).
22. C. S. Cheung, J. M. O. Daniel, M. Tokurakawa, W. A. Clarkson, and H. Liang, “High resolution Fourier domain optical coherence tomography in the 2 μm wavelength range using a broadband supercontinuum source,” Opt. Express 23(3), 1992–2001 (2015).
23. M. Maria, I. Bravo Gonzalo, M. Brandu, R. D. Engelsholm, T. Feuchter, P. M. Moselund, L. Leick, O. Bang, and A. Podoleanu, “A comparative study of noise in supercontinuum light sources for ultra-high resolution Optical Coherence Tomography,” Proc. SPIE 10056, 100560O (2017).
24. S. Pi, A. Camino, M. Zhang, W. Cepurna, G. Liu, D. Huang, J. Morrison, and Y. Jia, “Angiographic and structural imaging using high axial resolution fiber-based visible-light OCT,” Biomed. Opt. Express 8(10), 4595–4608 (2017).
25. S. Lawman, Y. Dong, B. M. Williams, V. Romano, S. Kaye, S. P. Harding, C. Willoughby, Y. C. Shen, and Y. Zheng, “High resolution corneal and single pulse imaging with line field spectral domain optical coherence tomography,” Opt. Express 24(11), 12395–12405 (2016).
26. N. Nishizawa, H. Kawagoe, M. Yamanaka, M. Matsuhashi, K. Mori, and T. Kawabe, “Wavelength dependence of ultrahigh-resolution optical coherence tomography using supercontinuum for biomedical imaging,” IEEE J. Sel. Top. Quantum Electron. 25(1), 1 (2019).
27. R. Leitgeb, C. Hitzenberger, and A. Fercher, “Performance of Fourier domain vs. time domain optical coherence tomography,” Opt. Express 11(8), 889–894 (2003).