Efficient sweep buffering in swept source optical coherence tomography using a fast optical switch

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Abstract: We describe a novel buffering technique for increasing the A-scan rate of swept source optical coherence tomography (SSOCT) systems employing low duty cycle swept source lasers. This technique differs from previously reported buffering techniques in that it employs a fast optical switch, capable of switching in 60 ns, instead of a fused fiber coupler at the end of the buffering stage, and is therefore appreciably more power efficient. The use of the switch also eliminates patient exposure to light that is not used for imaging that occurs at the end of the laser sweep, thereby increasing the system sensitivity. We also describe how careful management of polarization can remove undesirable artifacts due to polarization mode dispersion. In addition, we demonstrate how numerical compensation techniques can be used to modify the signal from a Mach-Zehnder interferometer (MZI) clock obtained from the original sweep to recalibrate the buffered sweep, thereby reducing the complexity of systems employing lasers with integrated MZI clocks. Combining these methods, we constructed an SSOCT system employing an Axsun technologies laser with a sweep rate of 100kHz and 6dB imaging range of 5.5mm. The sweep rate was doubled with sweep buffering to 200 kHz, and the imaging depth was extended to 9 mm using coherence revival. We demonstrated the feasibility of this system by acquiring images of the anterior segments and retinas of healthy human volunteers.

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1. Introduction

1.1. Motivation

The imaging speed of swept source optical coherence tomography (SSOCT) systems has increased dramatically in recent years, with A-scan rates in the megahertz regime recently being demonstrated for in vivo imaging [1–3]. These fast acquisition rates confer important advantages in numerous clinical applications. For example, high speed imaging of the ocular anterior segment minimizes artifacts due to patient motion, enabling improved accuracy in the extraction of biometric parameters [4–7]. High speed imaging is also important for Doppler OCT imaging of retinal blood vessels [8–10], where fast acquisition speeds facilitate the resolving of high speed flows in the optic disc without fringe washout [11]. Finally, and perhaps most critically, high speed imaging techniques are valuable for catheter-based intravascular OCT imaging, where the imaging time is limited by the requirement of administering saline flushes to remove blood from the OCT catheter’s field-of-view [12,13].

1.2. Sweep buffering in SSOCT

Sweep buffering in SSOCT refers to the use of long spools of optical fiber (typically hundreds to thousands of meters long) to delay and interlace the wavelength-tuning sweeps from SSOCT lasers. The first applications of sweep buffering in SSOCT were applied to Fourier Domain Mode Locked (FDML) lasers [14]. Because FDML lasers store a circulating copy of the laser sweep in a long fiber spool, the entire FDML cavity is, in a sense, a form of buffer. However, FDML lasers have also been demonstrated using two fiber spools inside the cavity to create unidirectional laser sweeps [15]. This is desirable as it has been observed that the “backward” sweep of an FDML laser (i.e. long wavelength to short wavelength) has superior noise and linewidth characteristics as compared to the “forward” sweep [14,15].

While the fastest OCT imaging speeds demonstrated to date have been achieved using swept source lasers, many of these lasers operate with duty cycles of less than 50%, including FDML lasers [14,15], fiber ring lasers [16,17] and external cavity tunable lasers (ECTLs) [18–20]. Sweep buffering can be used to increase the effective duty cycle of these lasers; this is achieved by splitting the laser output into two channels, delaying one channel by the sweep period and then recombining the two channels to produce a single interlaced channel [21]. This technique can also be multiplexed by constructing multiple buffering stages in series. Systems employing multiple buffering stages have been demonstrated with high speed, low duty-cycle FDMLs to realize up to 16-fold increases in imaging speed yielding A-scan rates as high as 5 MHz [1,2].

Unfortunately, these prior buffering techniques are relatively inefficient, as half of the light is lost at the terminal fiber coupler. This inefficiency can result in reduced imaging sensitivity, although for ophthalmic imaging applications this power loss only becomes important if the maximum power that can be directed to patient’s eye is below the ANSI limit [22]. It should be noted that the additional output from the buffer stage can also be used in a multi-spot configuration, as demonstrated in [1] and [21]. While this approach improves the effective imaging speed, it substantially increases system cost and complexity.

Furthermore, when imaging through birefringent tissues such as the cornea and retinal nerve fiber layer (RNFL), polarization mode dispersion (PMD) in the buffered output due to birefringence in the fiber spool can produce image artifacts and, for phase sensitive applications, corruption of the phase signal [23]. Baumann et al. [23] demonstrated that the artifacts created by PMD in the fiber spool could be avoided by using a linear polarizer after recombining the original and buffered sweeps. However, optical losses through the polarizer, coupled with the inefficiency of the buffered design, resulted in substantial optical attenuation, in addition to additional complexity and expense.

These optical power inefficiencies are often mitigated with the addition of a semiconductor optical amplifier (SOA) [1,2,23]. However, the use of an SOA adds costs and...
complexity, and may also negatively impact system performance. For example, in [23], the use of an SOA resulted in a narrower tuning bandwidth and broader linewidth, which degraded the axial resolution and axial imaging range, respectively.

An alternative method to mitigate the optical inefficiency of the buffering stage is to use a polarization-maintaining buffer design, as presented in [24]. This technique avoids the loss of half of the laser output at the terminal fiber coupler, and also precludes the creation of PMD in the laser output. However, the technique also results in alternating sweep types having orthogonal linear polarizations. While this is ideal for certain all-fiber implementations of polarization sensitive OCT [25], it may also result in the creation of artifacts in birefringent and diattenuating samples in conventional OCT.

Another consideration in the design of the buffering stage is related to the difference between the “sampling” duty cycle and the “laser-on” duty cycle of low duty cycle tunable lasers. Here, the sampling duty cycle refers to the proportion of the sweep period where the laser’s output wavelength is tuning linearly (or pseudo-linearly). By the laser-on duty cycle, we refer to the proportion of the sweep period during which the laser output power is at least 10% of its peak value. This often includes portions at the beginning and end of the laser sweep when the laser is not tuning, tuning slowly or reversing trajectory. In an ideal SSOCT laser, these two duty cycles would be the same, and the laser would not output appreciable power when not tuning optimally. Unfortunately, many commercially available SSOCT lasers, especially ECTLs, have different sampling and laser-on duty cycles, and unnecessarily expose the patient to optical power that does not contribute to the SSOCT signal. For the specific case of the commercially available ECTL used in the experiments described below, the sampling duty cycle was 46% but the laser-on duty cycle was 62%. Thus, if a 50/50 coupler were used at the end of the buffering stage, the beginning and end of the buffered sweeps would be corrupted by the end and beginning the original sweep, and vice versa. Between 0.8 μs and 1.2 μs of each 4.6 μs sweep is subject to this corruption, corresponding to between 17% and 26% of the sampling duty cycle. This crosstalk between the beginning and end of opposite sweep types results in an increase in shot noise, and also contributes to the ANSI limited power [22] that can be directed towards the patient eye. As a result of both of these effects, the system sensitivity would be reduced.

In this work, we describe a novel buffering strategy that employs a very fast optical switch (< 60 ns switch time) instead of a 50/50 fiber coupler at the end of the buffer stage. The use of this switch improved the efficiency of the buffering stage by over 70%, precluding the need for an SOA, which would have otherwise been required to achieve optimal sensitivity. The switch also prevented crosstalk between the beginning and end of opposite sweep types, further improving system sensitivity.

We also present a technique for transforming the wavenumber calibration signal from the original sweep signal (derived from the laser’s integrated Mach-Zehnder interferometer) for use with the buffered sweep, maintaining transform-limited axial resolution without requiring the construction of a second interferometer. In addition, we report on the results of a PMD analysis of the system, and devise a strategy to minimize polarization-related image artifacts without the use of lossy linear polarizers. Finally, we confirm that this technique is compatible with coherence revival-based heterodyning techniques [26,27], thus enabling simultaneous improvement of imaging speed and imaging depth in the same system. To demonstrate the feasibility of this technique for in vivo imaging, we acquired images of the anterior segments and retinas of healthy human volunteers.

2. Methods

2.1. Buffering stage design

A schematic of the buffering stage is shown in Fig. 1. The laser used was an Axsun Technologies ECTL with a central wavelength of 1040 nm and repetition rate of 100 kHz.
The total tuning bandwidth was 110 nm, 100 nm of which was tuned during the sampling duty cycle. Because the switch was operated in the dead-time between the original and buffered sweeps, the entire 100 nm bandwidth was conserved for both the original and buffered sweeps. The sampling and laser-on duty cycles were 46% and 62%, respectively. A 60/40 coupler was used to compensate the 1.5 dB attenuation (70% transmission) of the fiber spool such that both the original and buffered sweeps had similar power; this reduces the appearance of stripe artifacts in images combining both sweep types. The spool itself consisted of 1000 m of HI1060 fiber. Three polarization controllers were used to adjust the polarization at the input of the fiber spool and at both inputs to the optical switch. A commercially available optical switch produced by Boston Applied Technologies was used, based upon their Nanona FOS platform. The switch employs transparent electro-optic ceramics, which have been described previously [28], to produce a variable wave plate, which is then used to produce a polarization independent 2 x 2 optical switch. Specified parameters of interest of the switch were an insertion loss of 1.06 dB (78% transmission) and switching time of 60 ns, both of which were verified experimentally. The switch’s insertion loss compares favorably with that of 50/50 fiber couplers, which typically have insertion losses of 3.4 dB (excess loss of 0.4 dB). Thus, the use of the switch improves the buffer stage efficiency by over 70%.

By tuning the output voltage of the switch driver, the crosstalk between the output channels could be varied up to a maximum crosstalk suppression of 26 dB. Thus, the splitting ratio between ports could be adjusted to achieve any ratio from 99.7/0.3 to 50/50. By increasing the crosstalk, the second port of the switch could be used as an input to a MZI for wavenumber recalibration (k-clock). However, as we employed the integrated k-clock in the Axsun laser for these experiments, this port was not used, and thus the driver voltage was adjusted for maximum crosstalk suppression.

![Fig. 1. Schematic of buffering stage. PC: Polarization controllers. UP: Unused port.](image)

### 2.2. SSOCT system design

A schematic of the SSOCT system used to demonstrate the buffering system described above is shown in Fig. 2. The system made use of the spectrally balanced interferometer configuration suggested by Klien et al. [2]. A sample arm adapted for retinal imaging and a conventional reference arm were constructed. In compliance with the ANSI Z136.1 standard [22], power incident on the patient eye was limited to 1.8 mW. A 635 nm fiber coupled diode laser was also used as an aiming beam, but the power incident on the patient eye was limited to less than one microwatt (0.3% of the ANSI-limited maximum permissible exposure) and thus did not pose an appreciable exposure hazard. OCT light returning from the reference and sample arms was detected with a NewFocus 1807 balanced receiver, an InGaAs receiver with 120 MHz electronic bandwidth. An Alazar Technologies ATS9870, operating at 1 GS/s was used to record both the integrated k-clock signal from the Axsun laser and the interferometric signal from the balanced receiver.
To demonstrate the feasibility of combining sweep buffering with coherence revival [26], we modified the SSOCT system presented above to enable extended depth imaging of the anterior segment. Coherence revival was used in a + 1 cavity length offset configuration (i.e. with the sample arm one cavity length longer than the reference). Other changes to the system included replacing the NewFocus balanced receiver with a Wieserlabs receiver (WL-BPD1GA) which had substantially higher electronic bandwidth (1 GHz). Due to the low transimpedance gain of the Wieserlabs receiver, an RF amplifier (HD24388, HD Communications Corp.) was also used. Incident power remained at 1.8 mW.

2.3. Hardware triggering and timing

The delay between the original sweep and buffered sweep was calculated by measuring the outputs from laser and the spool simultaneously on two separate photoreceivers. The peak of the cross-correlation of these two signals was taken to represent the optical delay in the fiber, which was measured to be 4.911 μs. This measurement was repeated three times and found to be stable within the resolution of the measurement (1 ns).

A schematic of the data used to perform this cross-correlation is shown in Fig. 3. The solid black trace represents the original signal while the dashed gray trace represents the buffered sweep. For the purposes of data processing, the original sweep was defined to occur between $t = 0$ and $t = 4.608$ μs, and the buffered sweep was defined to occur between $t = 4.911$ μs and $t = 9.519$ μs. In Fig. 3, the beginnings of the original and buffered sweeps are denoted by dashed green and dash-dot magenta vertical lines, while the end of the sweeps are denoted by dotted red vertical lines. This figure also demonstrates the appreciable optical crosstalk that would occur between the two sweeps if a 50/50 coupler were used.

![Fig. 3. Traces of original (solid black) and buffered (dashed gray) sweeps.](image-url)
The hardware triggering of this system was slightly more complicated than that of a traditional SSOCT system, and was carried out as follows. First, because the laser only fired a single start trigger for each original sweep/buffered sweep pair, the digitizer was used to acquire a continuous stream of 9536 samples (9.536 μs) for each sweep trigger, which thus contained both the original and buffered sweeps. The digitizer thus operated at a 95% duty cycle, and output a digital high (5 V) on its auxiliary IO (auxIO) channel when acquiring. The rising edge of this auxIO channel was used to trigger a Stanford Research Systems (SRS) DS345 digital function generator, which was set to output a single square wave pulse at a frequency of 107.3 kHz and initial phase of 22 degrees. This effectively produced a square wave that was low for the first half of the sweep period and transitioned to high just before at the start of the buffered sweep. This square wave was used as the control signal for the optical switch driver. Finally, a TGP110 pulse generator (Aim-TTi) was synchronized with the SRS function generator and set to create a doubled pulse for each line trigger, with an inter-pulse delay of 4.6 μs. The output from the pulse generator was then fed into a National Instruments DAQ card (NI PCI-6221) and used as the sample clock for the drive waveform for the galvanometers. A schematic of the timing scheme is shown in Fig. 4.

![Fig. 4. Timing and triggering signals in the buffered SSOCT system.](image)

2.4. Management of polarization mode dispersion

Both the fiber spool and the optical switch result in the generation of PMD, which in turn may result in the creation of image artifacts and ghosts when imaging a birefringent sample. PMD occurs in the fiber spool because “single mode” fibers actually support two orthogonally polarized modes. Bending stresses induced by wrapping around the spool induce birefringence in the fiber, resulting in a differential group delay between the two polarization modes [29].

The generation of PMD in the optical switch occurs by a different mechanism. As described above, the optical switch functions by splitting incoming light into two polarization channels. An electro-optic device is then used as a tunable waveplate to impart either a 0 or 90 degree rotation of the polarization state in each channel. Two polarization channels are then combined and directed into either of the two output ports. As a result of the design of this
The combination of PMD from the spool and from the switch complicates matters further, especially if the fast and slow axes of the two components are not aligned. If such a system were used in OCT, an A-scan of a point reflector (i.e. a mirror) would result in not one, but several peaks corresponding to the interference between differentially delayed pulses in the reference and sample. If this system were used for imaging, its axial point spread function (PSF) would also have many peaks, resulting in severe image artifacts.

Fortunately, the PMD in this system can be managed through the use of fiber polarization controllers. For the buffered sweep, because the laser is partially polarized [27], we used a polarization controller at the entrance of the spool to align the polarization axis of the laser output to one of the polarization axes of the spool. A second polarization controller was used to align the polarization of the light leaving the spool to the axes of the switch. Similarly, for the original sweep, a polarization controller was placed before the switch to align the polarization axis of the laser output to one of the axes of the switch. While these polarization alignments cannot be perfect due to chromatic considerations, they significantly reduced the effects of PMD by largely restricting power in the system to only two of the four possible polarization states. Furthermore, these two dominant polarizations are orthogonal, and thus, in the absence of birefringence in the sample arm, each polarization state should only interfere with itself. This would result in a single peak, analogous to an OCT system without PMD.

The final consideration in polarization management is the potential presence of birefringence in the sample. By placing a polarization controller in the reference arm, birefringence can be introduced into the reference arm that offsets the birefringence in the sample arm. Thus, when imaging through birefringent tissue, such as the cornea or the RNFL, the reference arm polarization must be optimized. As we demonstrate below, these polarization artifacts could generally be suppressed beneath the noise floor with careful adjustment of these polarization controllers.

To demonstrate the effects of these multiple polarization states, we performed a series of experiments using various configurations in the OCT source arm, as depicted in Fig. 5. First, we constructed an OCT system with the switch placed in the source arm, but without the fiber spool (Fig. 5A). Second, we replaced the fiber spool with the switch (Fig. 5B). Finally, we took measurements with both the spool and switch in the source arm (Fig. 5C). For each experiment, we acquired A-scans with only a mirror in the sample arm, and deliberately misaligned the polarization controllers to enable visualization of the multiple peaks corresponding to interference between the various polarization states.

The results of these measurements are shown in Fig. 6. To aid in comparison, the A-scans shown have been normalized to have the same integrated amplitude. Figure 6A shows an A-scan of a mirror with only the switch in the source arm, and contains three peaks each separated by approximately 40 microns. Figure 6B shows an A-scan of a mirror with only the
spool in the source arm. Because the PMD through the spool is comparable to the axial resolution of the system, the expected three peaks overlap partially. Thus, the reference arm polarization controller was adjusted to emphasize only the outer two peaks. These two outer peaks were separated by about 16 microns, implying that the separation between individual peaks was 8 microns, 5 times less than the peak separation through the switch. Figure 6C shows an A-scan of a mirror with both the spool and the switch in the source arm, and shows 3 pairs of split peaks. Each pair of peaks corresponded to the 3 peaks created by PMD in the switch, and the centers of the pairs were separated by approximately 40 microns, as expected. The splitting of the peaks was due to the PMD in the spool, and the separations between the paired peaks were all approximately 16 microns, as expected. Finally, Fig. 6D shows the axial PSF of the system in the same configuration as in Fig. 6C, but after the polarization controllers were optimized to maximize the central peak. While some residual sidelobes remain, the largest sidelobe peak is approximately \(-42\) dB weaker than the central peak. For all of these measurements that used the fiber spool, wavenumber recalibration was performed as described below.

2.6. Wavenumber recalibration and phase stabilization

To facilitate wavenumber resampling of both the original and buffered sweeps, the signal from the integrated k-clock of the Axsun laser was digitized and processed as follows. For real-time display, only the forward sweeps were displayed. The zero-crossings of the clock were detected and used to generate a linear-in-wavenumber recalibration vector. This recalibration vector was interpolated such that the vector length matched the number of data samples acquired during the original sweep (4608), which provided \(z_{\text{max}} = 12.4\) mm. This interpolated vector was then used to resample the portion of the photoreceiver signal corresponding to the original sweep to be linear in wavenumber.
In post-processing, wavenumber recalibration of the original sweep was performed using a phase linearization algorithm, as described previously [30,31]. By appropriately modifying these recalibration algorithms, the buffered sweep could also be resampled. Dispersion in the fiber spool causes short wavelengths to travel slower than long wavelengths, and because the laser sweeps from short wavelengths to long wavelengths, the entire laser sweep is compressed after travelling through the spool. If we were to direct this compressed laser sweep into an MZI and compare the resulting signal with the MZI signal from the original sweep, we would expect the zero-crossing intervals to be progressively shortened (i.e., the fringes to be chirped) towards the end of the laser sweep. Thus, by adjusting the timings in the resampling vector for the original sweep, we were able to estimate what the resampling vector would be for the buffered sweep. A description of this transformation follows.

The linear-in-wavenumber timing vector derived from either the zero-crossing or phase retrieval algorithms was modified by applying a polynomial delay function as follows:

\[
T_2[n] - \text{delay} = T_1[n] + b_0 n + b_1 n^2 + b_2 n^3 \ldots
\]

Here, \(T_2\) is the linear-in-wavenumber timing vector for the buffered sweep, \(n\) is the index of the zero-crossing or phase retrieval measurement, \(\text{delay}\) is the delay between the original and buffered sweeps (4.911 μs), \(T_1\) is the linear-in-wavenumber timing vector of k-clock, and the \(b\) coefficients are fitting parameters. This technique is very similar to the dispersion compensation techniques described in [26] and [32]. Thus, using an A-scan of a mirror, an optimization approach was used to find the \(b\) coefficients, similar to the approach described in [26] to find dispersion compensation coefficients. However, to ensure that the fitting of the \(b\) coefficients is correcting for wavenumber recalibration only and not dispersion, it is important to minimize dispersion between the mirrors in the sample and reference arm, or to account for the dispersion in the optimization. This latter technique works because the dispersion parameters should be identical between the original sweep and the buffered sweep. Using this technique, we were able to achieve identical sensitivity, roll-off and axial resolution between the original sweep and buffered sweep using only two \(b\) coefficients (\(b_0\) and \(b_1\)).

Phase stabilization was also performed in post-process to reduce the appearance of fixed pattern noise. In general, stabilization is achieved by precisely aligning the spectral interferograms in wavenumber before Fourier transformation [33]. While we have previously demonstrated that this alignment can be performed in real time with the use of an external wavelength reference [11], in this work we only applied phase stabilization in post-process, using a technique similar to that described in [33]. Briefly, the relative shift between spectral interferograms was measured by cross-correlating the k-clock signal from the first A-scan with that of each subsequent A-scan. This algorithm was applied separately to the A-scans corresponding to the original and buffered sweeps of each B-scan. The relative shifts were corrected by appropriately shifting the wavenumber recalibration vectors. DC subtraction was then performed prior to Fourier transformation to remove fixed pattern noise. The presence of the fiber spool did not appear to negatively impact phase stabilization, as both the original sweeps and buffered sweeps exhibited comparable fixed pattern noise suppression.

2.7 In vivo imaging

To demonstrate the feasibility and performance of this buffered SSOCT system, we acquired images of the posterior eye from healthy human volunteers. To manage PMD, the system was optimized as above with a mirror placed in the reference arm. Then, during patient alignment, only the reference arm polarization controller was adjusted while observing bright layer in the resulting image (usually the RNFL). The polarization was adjusted until this layer appeared as bright and sharp as possible with minimal ringing artifacts. B-scans were then acquired with 1600 A-scans per B-scan at a frame rate of 125 Hz.
To demonstrate the feasibility of combining sweep buffering with coherence revival, the system was modified to enable extended depth imaging of the anterior segment. Reference arm polarization was optimized until minimal ringing artifacts were observed and B-scans were then acquired as before.

3. Results

3.1. Peak sensitivity and sensitivity fall-off

The system’s peak sensitivity and sensitivity fall-off were measured experimentally and compared to the theoretical shot-noise limited performance. For the standard configuration, sensitivity was measured at a depth of approximately 500 μm and found to be 97 dB with 1.8 mW on the sample, as compared to a theoretical sensitivity of 102 dB. The 5 dB discrepancy is attributed to coupling losses, residual RIN and digitization noise. In the coherence revival configuration, peak sensitivity was located at a depth of approximately 6 mm and found to be 94 dB with 1.8 mW on the sample. The additional 3 dB discrepancy in this configuration is attributed to the coherence revival sensitivity penalty [26], and increased noise from the alternate receiver and amplifier.

The fall-off performance in both configurations were identical to the results previously reported in [27]. A 6 dB imaging range of 5.5 mm was observed for the standard configuration, and a 9 mm imaging range was observed for the coherence revival configuration. Both the peak sensitivity and 6 dB fall-off distance were identical for the forward and buffered sweeps, in both the standard and coherence revival configurations.

3.2. Imaging results

Representative B-scans of the foveal region of the retinas of two healthy human volunteers are shown below. Each frame consisted of 800 A-scan pairs, for a total of 1600 A-scans. Acquisition time for each frame was 8 ms (corresponding to 125 Hz), and thus these 5 x averaged images represent the image quality achievable at “video rate” (25 Hz).

Figure 7 shows the typical image quality achieved before fine adjustment of the reference arm polarization controller (but after optimizing the controller on a mirror reflector). Some ringing artifacts can be seen above the RNFL, but these artifacts are much weaker than the peak signal, and do not appear surrounding the dimmer layers. Adjustment of the reference polarization controller can suppress such artifacts beneath the noise floor, as shown in Fig. 8.

Figure 9 shows a representative B-scan of the anterior segment of a healthy human volunteer acquired with the coherence revival configuration. As before, each B-scan consisted

![Image]

Fig. 7. In vivo B-scan of fovea acquired before optimization of polarization controller. Image consists of 1600 A-scans x 5 averaged frames. Total acquisition time was 40 msec. Scale bars are 250μm (vertical) and 1° (horizontal). Arrows point to artifacts due to residual polarization mode dispersion.
Fig. 8. \textit{In vivo} B-scan of fovea acquired after optimization of polarization controller. Image consists of 1600 A-scans x 5 averaged frames. Total acquisition time was 40 msec. Scale bars are 250\micro m (vertical) and 1° (horizontal).

Fig. 9. \textit{In vivo} B-scan of anterior segment. Image consists of 1600 A-scans and x 5 averaged frames. Total acquisition time was 40 msec. Scale bars are 1 mm. Inset: 300\% enlargement of the apex of the cornea of 800 A-scan pairs (1600 A-scans) and was acquired in 8 ms, supporting a frame rate of 125 Hz. The artifact in the anterior chamber of this image is a lens reflection that appeared due to coherence revival. Although not performed on this image, lens reflection artifacts can be removed by background subtraction. The inset in Fig. 9 shows an enlargement of the apex of the cornea, demonstrating the absence of ringing artifacts due to PMD. The subtle ringing that is observed at the apex of the cornea is due to detector saturation from the bright specular reflection.

4. Discussion

We have demonstrated the development of a novel buffering strategy for increasing the imaging speed of SSOCT systems that operate with “sampling” duty cycles of 50\% or less. This technique has numerous advantages over previously described buffering techniques. First, it is very power efficient, with over 60\% of the source power being transmitted through the buffering stage. This represents more than a 70\% improvement over configurations employing a terminal fiber coupler. It also exhibits superior sensitivity because the patient eye is not exposed to light that is not used for imaging at the tails of the SSOCT sweep. Lastly, the system employs numerical methods that allow the use of the original sweep’s calibration signal for recalibration of the buffered sweep. This dramatically simplifies the implementation of buffering with turnkey systems that contain integrated k-clocks.
In addition, the buffering system described here is compatible with coherence revival-based heterodyne SSOCT described in [26]. The combination of these techniques resulted in an SSOCT system with twice the imaging speed and nearly twice the imaging depth of conventional SSOCT systems employing the same source.

While the presence of PMD in the source arm can result in the generation of image artifacts, we have demonstrated that these artifacts can be mitigated by careful manipulation of fiber polarization controllers. For example, the image in Fig. 8 shows sharp boundaries between retina layers that are not corrupted by artifacts. Similarly, the inset in Fig. 9 shows the bright reflection at the air-cornea boundary that also does not exhibit any visible ringing artifacts (except for saturation).

In summary, we have established that the presently described buffering technique is compatible with coherence revival-based heterodyne SSOCT, and demonstrated the simultaneous improvement of both the imaging speed and imaging depth of an SSOCT system based upon a turnkey, commercially available laser.

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