Long-range and wide field of view optical coherence tomography for in vivo 3D imaging of large volume object based on akinetic programmable swept source

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Abstract: Current optical coherence tomography (OCT) imaging suffers from short ranging distance and narrow imaging field of view (FOV). There is growing interest in searching for solutions to these limitations in order to expand further in vivo OCT applications. This paper describes a solution where we utilize an akinetic swept source for OCT implementation to enable ~10 cm ranging distance, associated with the use of a wide-angle camera lens in the sample arm to provide a FOV of ~20 x 20 cm². The akinetic swept source operates at 1300 nm central wavelength with a bandwidth of 100 nm. We propose an adaptive calibration procedure to the programmable akinetic light source so that the sensitivity of the OCT system over ~10 cm ranging distance is substantially improved for imaging of large volume samples. We demonstrate the proposed swept source OCT system for in vivo imaging of entire human hands and faces with an unprecedented FOV (up to 400 cm²). The capability of large-volume OCT imaging with ultra-long ranging and ultra-wide FOV is expected to bring new opportunities for in vivo biomedical applications.

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1. Introduction

First introduced in 1991 [1], optical coherence tomography (OCT) has been amply demonstrated as a powerful inspective tool due to its capability of depth-resolved imaging with high resolution [2]. However, most of current OCT applications suffer from small area of picturing within relatively flat and shallow layers due to its relatively short ranging distance and small field of view (FOV). Efforts to expand the imaging FOV have demonstrated a great value in peripheral retinal examination [3], full eye assessment [4], whole brain vascular visualization in neuroscience [5], and skin imaging [6]. In many biomedical applications, depth-resolved imaging of large biological sample with wide FOV (> 10 cm²) is often necessary, as proven by other imaging techniques such as X-ray computed tomography, ultrasound imaging and magnetic resonance imaging. There is also a demand in the market of manufacturing industry where non-destructive and precise visualization of the sample over a large area is often required. Thus, it would be of great value if OCT can be developed to have extended imaging range and increased FOV.

In the early times of time-domain OCT, the cross-sectional information was resolved by scanning the reference mirror, limiting the system’s imaging speed and sensitivity. These limitations have been mitigated to a large extend since the advent of Fourier-domain OCT (FD-OCT) [7]. As one of the main categories in FD-OCT, spectral-domain OCT (SD-OCT) becomes the main stay for the most commercialization efforts for applications in ophthalmology and endoscopy. However, the spectral resolution in SD-OCT is determined by the grating, beam size and line-scan camera in the spectrometer, which unfortunately limits the ranging distance for most SD-OCT systems from ~2 to 6 mm. While the depth ranging can be extended by the use of full range technique [8], the maximum ranging distance is still limited to less than 16 mm for useful clinical applications [9]. In addition, SD-OCT system is somewhat cumbersome due to its bulky size of the free-space configuration in spectrometer for detecting the spectral interferogram. Moreover, the A-scan rate of SD-OCT is...
fundamentally limited by the speed of commercially available camera, which is typically between ~100 and 200 kHz for the state-of-the-art high-speed line-scan cameras.

Another branch of FD-OCT is based on the technique of swept laser source. Swept-source OCT (SS-OCT) [10] generates wavelength-scanning light with much narrower instantaneous linewidth, avoiding the use of bulky spectrometers in SD-OCT. Combined with high-speed photodetector and data acquisition, SS-OCT is proven capable of achieving much better spectral resolution, leading to greatly increased system ranging distance when compared to its SD-OCT counterpart. In addition to its long imaging range, SS-OCT is also promising to offer other advantages, including faster A-scan rate, reduced fringe wash-out effects due to motion artifacts from sample or scanner, compact system configuration and improved detection sensitivity [4].

In recent years, there is a surging of new techniques in the development of suitable swept laser sources for OCT imaging. The early SS-OCT was demonstrated in the late 1990s with a sweeping rate of merely 10 Hz by scanning a galvanometer in grating-tuned external cavity [11]. The fast scanning techniques by the use of resonant galvanometer and rotating polygon mirror have since been developed that enabled the sweeping rate up to 10s-100s kHz [12, 13], where the tuning configuration has relatively long resonators due to its bulk optics or fiber components in the external laser cavity. Such configuration, however, generates multiple longitudinal modes and requires relatively long time to sweep the laser. Thus, the coherence length is limited to within several millimeters, which is not ideal for imaging applications where large-volume samples are the target of interest. The technique of Fourier-domain mode locking (FDML) uses a fiber ring swept laser, in which an intra-cavity tunable filter is utilized to fast scan the wavelengths [14]. With optimized configuration such as dispersion compensation, FDML is shown capable of extending the coherence length to > 21 mm [15]. Impressive ultrafast sweeping rate of up to 5.2 MHz has been demonstrated by multiplying the fundamental A-scan rate of FDML with optical fiber buffering [16]. Vertical-cavity surface emitting laser (VCSEL), combined with micro-electromechanical systems (MEMS) technology, greatly reduced the laser cavity length and achieved single longitudinal mode operation, giving very narrow instantaneous linewidth [17]. With VCSEL, a superior coherence length of > 100 mm has been demonstrated. However, all of the aforementioned swept sources are realized by mechanical-sweeping mechanism, which unfortunately provides an inherent limitation of accumulation and depletion of momentum, leading to hysteresis and unstable drifting that affects the stability and life time of the light source [18]. To mitigate the problem related to the momentum, non-mechanical sweeping mechanism is desirable. In response to this request, fast electro-optical deflector driven by electrical voltage in an external-cavity laser has been proposed that demonstrates a sweeping rate of 200 kHz A-line rate [19]. This electrical controlled operation possesses a high phase stability with low trigger jitter, and is expected to achieve higher sweeping rate than mechanical sweeping mechanism. The technique of optical time-stretch provides ultra-fast swept rate by a passive sweeping mechanism, which has been demonstrated to deliver multi-megahertz A-scan rate [18, 20].

Another promising non-mechanical swept laser is akinetic all-semiconductor programmable swept source [21]. This is an electrically driven swept source based on Vernier-tuned distributed Bragg reflector (VT-DBR). The rapid sweeping of laser wavelength is achieved by changing the refractive index of DBR mirrors in the laser cavity through the control of the current. The akinetic swept source has been integrated within a single chip using semiconductor opto-electronic design, making the device extremely compact and cost-effective. Benefiting from the akinetic design without any mechanically-moving parts and with programmable control of its sweeping speed, the akinetic swept source has demonstrated superior phase stability and sweep repeatability [21]. In addition, the short cavity length in the compact laser chip gives rise to only one longitudinal mode, thereby leading to extremely long coherence length. These advantages of this akinetic swept source provide opportunities
for developing an OCT system that can deliver ultra-long ranging distance and ultra-wide FOV imaging.

In this paper, we report a high-speed ultra-long range and ultra-wide FOV SS-OCT system by exploring the akinetic all-semiconductor programmable swept source. The single longitudinal mode operation over a broad spectral bandwidth of the swept source enables long coherence length with high resolution. Moreover, an adaptive calibration approach is proposed to compensate tiny non-linearity in optical frequency (k-space) from the laser source sweeps, which otherwise would affect the imaging performance when dealing with the target located at far distance. This procedure dramatically improves the OCT point spread function (PSF) performance when sampling frequency is required to be much higher than the factory default settings in order to provide uncompromised sensitivity over the extended depth range. With the implementation of ultra-wide angle optical and mechanical configuration and high-speed data acquisition, we demonstrate unprecedented long-range and wide-field OCT system for large-volume in vivo imaging.

Fig. 1. Schematic of the experimental setup for UW-OCT system. Lower left insert indicates measured spectrum of the akinetic light source running at 100 kHz.

2. Experimental implementation

2.1 OCT imaging system design

The experimental setup of the Ultra-long ranging and ultra-Wide FOV OCT (UW-OCT) system is shown in Fig. 1. In this setup, we employed an akinetic all-semiconductor programmable swept source commercially available from Insight Photonic Solutions, Inc. The light source provides a high-speed sweep rate of 100 kHz centered at 1300 nm with a top-hat spectrum of 100 nm bandwidth (see lower left insert in Fig. 1). The working principle and basic performance has been described in detail in [21]. The sweep mechanism is achieved by a full electronic control of laser operation without any moving parts, potentially overcoming most of the limitations encountered with mechanically sweeping lasers and providing much improved sweeping linearity with single longitudinal mode operation. The swept laser output was coupled into fiber-based Mach–Zehnder interferometer via broadband fiber couplers with different coupling ratio and broadband optical circulators (as shown in the
A pair of XY galvanometric mirrors (GVS002, Thorlabs Inc, USA) was used in the system to raster-scan the beam spot over the sample in order to obtain 3D volumetric imaging of the sample. To achieve wide FOV imaging, we employed a wide-angle camera lens in the sample arm with 290 mm focus length, producing spot sizes of ~160 μm and focus depth of ~62 mm to provide a FOV of up to ~20 x 20 cm². With ~5 mW incident light power on the sample and 30 dB neutral density filter (60 dB both ways in the OCT configuration), the system sensitivity was measured to be as high as ~105 dB for OCT imaging. The back-reflected lights from the reference and sample arms were coupled back into the fiber and interfered with each other after the combination by the 50:50 fiber coupler. In order to provide long ranging distance, a high-speed balanced photo-detector (1.6 GHz bandwidth, Thorlabs Inc, USA) was used to detect the OCT interferograms.

2.2 Large volume data acquisition and management

A high-speed digitizer (ATS9373 AlazarTech Inc., Canada) was employed to record the spectral interference fringe signals, providing 12-bit resolution with up to 4 GS/s sampling rate, enabling a maximum depth as long as ~12.5 cm in our system. Although the full speed of the digitizer is 4 GS/s, the maximum continuous sampling speed was limited to 3.2 GS/s, due to the bandwidth of the PCIe 3.0 x8 interface employed in our system, which has a maximum data throughput at 6.8 GB/s. In continuous capturing mode, the raw data was streamed to a high speed, solid-state hard-disk array in RAID0 configuration to avoid buffer overflow. All of the operations, including laser sweep control, galvo-scanning, data acquisition, transmission and storage, were automatically controlled by the host computer using a software package written in LabVIEW. This all-programmable UW-OCT system provides a convenient and efficient platform for ultra-long range and ultra-wide field imaging.

2.2 Adaptive calibration for spectral over-sampling

Based on the operating principle of akinetic swept laser, the sweep of laser cavity states does not generate a subsequent and contiguous ensemble of valid wavelength sweep points [21].
During the self-calibration of laser source, the valid wavelength points are recorded and the “data valid vector (DVV)” is generated accordingly. In the OCT image reconstruction procedure, DVV is applied as a mask to recover the entire wavelength sweep. By factory default, the laser sweep repetition rate is 96.15 kHz. With the default settings of 400 MHz sampling frequency, the number of total sweep points is 4160, including 2878 valid points. As a consequence, the imaging depth range is limited to ~12.5 mm. In order to achieve a depth ranging beyond this limit, the sampling data points per sweep must be increased. One way to accomplish this is to slow down the sweep rate, since the Insight akinetic swept laser is fully programmable. In this way, if the sweep is slowed down to 20 kHz, the image depth ranging can be extended to 5 times that of the factory setting [21]. However, reducing the sweep rate would inevitably slow down the imaging speed, leading to the data acquisition forbiddingly long, not desirable for most in vivo imaging applications. Our strategy to increase the ranging distance is to speed up the data acquisition during each wavelength sweep, i.e. oversampling the spectrum while not sacrificing the sweeping speed of the laser. With the full speed of the state-of-art AlazarTech digitizer, the 4 GS/s speed provides 10x oversampling of the spectrum. In doing so, the imaging depth range can be potentially extended to 12.5 cm, with the sampling data points of 40,000 per sweep. Due to the limitation of data transfer as mentioned in the last section, the data acquisition was limited to 3.2 GHz in our study, which provided a ranging distance of 9.5 cm. Consequently, the DVV is scaled accordingly, to ensure the integrity of wavelength sweep.

While it makes sense to use faster data-acquisition to over-sample the spectrum to achieve longer depth ranging, the spectral over-sampling rapidly deteriorates the performance of OCT point-spread-function (PSF) at the extended depth. This is due to the tiny non-linearity of wavelength sweeps from the laser. While the wavenumber-time (k-t) relationship is linear exactly at 400 MHz sampling points with a sweep-to-sweep linearity deviation less than 0.002% [24], the oversampled points are slightly away from the perfect straight line, due to non-linear transition of the laser states between 400 MHz time points. With a sampling rate at 3.2 GHz, Fig. 2 shows a measured PSF roll-off curve over 9.5 cm ranging distance without calibration, where the PSF is almost flat due to the k-t linear relationship for the frequencies up to its internal clock of 400 MHz in the Insight light source. However, the PSF rolls off quickly after the optical delay exceeding 400 MHz, coupled with rising side-lobes around the main peaks. Both these effects limit the PSF performance beyond 2.5 cm optical delay, with a measured peak-to-side lobe ratio (PSR) less than 25 dB on the lower half of the imaging depth range (> ~4.5 cm). This would affect the eventual OCT imaging performance if the object is located at a ranging distance of > 2.5 cm. Figure 2 also gives the resolution performance measured at the full width half maximum (FWHM) of the PSFs, where the gradual deterioration of axial resolution is observed, i.e., the resolution starts to become worse once the object is located beyond 2.5 cm, and deteriorates rapidly after 6.0 cm optical delay.

In order to overcome the limitation of the PSF roll-off and improve the overall imaging sensitivity over the entire ranging depth, below we introduce an adaptive calibration procedure in the post-processing of OCT signals. This calibration method is a refinement of the conventional calibration method that are commonly used for correcting k-space non-linearity in FD-OCT. First of all, prior to data acquisition, a series of reference interferograms were captured with a series of optical delays that are evenly spaced to cover the whole imaging depth range, upon which the unwrapped instantaneous phases (Hilbert phase) at a specific optical delay were then retrieved. This process is identical to the conventional approach, except that here the calibration is repeated for multiple depth/optical delays. This series of calibration data forms a k-space calibration table. Each element in this table is a curve for resampling the spectrum to correct for the non-linearity of k-space, specifically for a designated optical delay band. Linear interpolation is used to perform resampling. A more detailed elaboration of some special considerations is given below.
In Fig. 3(a), three measured PSF curves around 480 MHz (~3 cm) are shown, without any calibration, where the peak-to-side lobe ratio is seen about 30 dB. At this depth, the side-lobes are likely to deteriorate the image quality since they start to appear higher than the noise floor. After using the calibration data at 480 MHz from the look-up table for k-linearization, all three PSFs are significantly improved, with peak-to-side lobe ratio higher than 45 dB, as shown in Fig. 3(b). Some notable side-lobes also arise after calibration, at the frequencies of Fp ± 480 MHz where Fp stands for the main-peak frequency, which were likely due to the side-effect of the calibration. However, these side-lobes are far from the main peaks (~3 cm away), so they would have no effect on the OCT imaging performance because the actual useful signals are sparsely presented in the captured data sets due to the fact that the penetration depth of OCT imaging of biological tissue is typically less than 3 mm. It is also notable in the PSF before calibration that some ghost peaks are present at Fp ± 800 MHz and Fp ± 1600 MHz. These are the artifacts of the electronics since the 400 MHz laser driving clock gives harmonics at 800 MHz. Similar to the side effects in the calibration, those ghost lines would not have notable effects on the actual OCT imaging performance.

Although the above single-depth calibration method works for its adjacent frequency bands, the performance over the whole ranging distance is unfortunately not optimized due to
the tiny non-linearity between the factory 400 MHz sweeping points. Figure 3(c) gives the PSF measurements over the entire depth of 9.5 cm when the calibration was only performed by the calibration data at the 480 MHz depth position, where it can be seen that this calibration procedure only works within ± ~50 MHz bands around the calibration frequency (480 MHz) and its harmonics (960 and 1,440 MHz) (Fig. 3(e)). At other frequencies, this single depth calibration would actually deteriorate the performance. This effect is shown in Fig. 3(f), where the PSFs were measured at three frequencies around 720 MHz. Such deteriorating effect is likely due to a combined effect of the periodicity of k-linearity errors, and Hilbert phase errors when generating the calibration data. Figure 3(d) gives the PSF measurements over the entire depth when the calibration was performed at the frequency of 780 MHz, where as it is expected, the calibration is only effective at the 780 MHz and 1,560 MHz bands.

Our solution to mitigate the above problem for ultra-long ranging applications is to use an adaptive calibration procedure in which the calibration is applied to the spectrum data set adaptively according to the actual position of the target using the calibration look-up table. Because of harmonics effect in the calibration as shown above, there may be a number of ways to build an optimal look-up table. Here, we just provide a simple example, where we performed the calibration at the pre-set depth positions at 180, 280, …, 780 MHz with an interval of 100 MHz. This look-up table is saved for later use during practical imaging. As such in practice, the calibration data at a specific depth is only applied onto the spectrum data set so that the imaging performance for the target is kept optimum. This is feasible because of the fact that the penetration depth of OCT imaging of biological tissue is typically less than 3 mm, which corresponds to a frequency bandwidth of ~50 MHz. Because of the harmonics effect as shown above, practically one would only need to perform the calibration for the frequencies up to 0.25 Fs (sampling frequency). For example, for the depth band around 1,560 MHz at the bottom of the image, the calibration data at 780 MHz would work equally well.

Fig. 4. Sensitivity roll-off and resolution measurement of the a kinetic SS-OCT system, after adaptive calibration.

To use the look-up table in practice, it is essential to know the rough depth position of the target, upon which the calibration data set from the look-up table can then be applied to arrive at the final results. While straightforward, such approach would demand twice the computational power to achieve the final results because FFT has to be applied twice for each
A-lines, i.e., one before and one after the calibration. One way to improve this situation and speed up the processing is to use only partial FFTs around the target depth position because of the fact that the actual useful OCT signals are sparse in the captured interferograms. In doing so, the depth position of the target can first simply be determined by processing a 1/16 fraction of the acquired OCT spectrum, which provides the rough position of the target. This approximate position is then used to select the calibration data from the look-up table to calibrate the captured interferogram. Thereafter, to further save the computing power, a fractional discrete Fourier transform algorithm [22] is applied to the calibrated interferogram to obtain only a segment of the entire depth, centered at the depth position with a bandwidth of 100 MHz, i.e. only a range of ~6 mm signals are computed. In this way, the computational cost is significantly reduced compared to that performing FFT of the entire depth, while keeping the fidelity of OCT image representing the target.

3. Results

3.1 Assessment of OCT system performance after adaptive calibration

To assess the system sensitivity performance over the entire ranging distance after the adaptive calibration procedure described in the last section, a mirror was installed in the sample arm, and then the roll-off curve of the system point spread function (PSF) was measured by varying the optical path length in the reference arm. The results are shown in Fig. 4, where we used a look-up table that consisted of 7 calibration data sets obtained from 180, 280, ..., 780 MHz with an 100 MHz interval. Figure 4 also gives the measured full width half maximum of the PSFs against the ranging distance, where the mean value of resolution is 13.4 μm, with maximum value of 15.6 μm, which is close to the theoretical axial resolution of
From the results, the sensitivity roll off is less than 3 dB over a long distance of 4.75 cm. Comparing the sensitivity roll-off before and after adaptive calibration, −10 dB roll-off distance is improved from 5.7 cm to 6.9 cm. The total roll-off of sensitivity is 14 dB over 9.5 cm, and the peak-to-side lobe SNR maintains higher than 30 dB across the most of the imaging depth range. This unprecedented roll-off performance of our OCT system is largely due to the single mode operation without mode hop in the akinetic swept laser, together with the adaptive calibration procedure as we described in the last section. Please note that the coherence length of the light source is actually in excess of the maximum depth we could measure with our digitizer operating at 3.2 GS/s.

To assess the OCT image quality after employing the adaptive calibration procedure, we captured the 2D cross-section images (shown in the top row of Fig. 5) of a semi-transparent roll of tape at different depth positions. By varying the optical path lengths in the reference arm, the depth position of the tape roll was changed from a position almost close to zero delay line up to a maximum ranging distance allowed by the system. For comparison, the same results without calibration procedure are also shown in the bottom row of Fig. 5. From Fig. 0.5, it can be seen that the calibration procedure is effective for the target sample that is located at the depth positions beyond ~2.5 cm, agreed well with the findings in the last Section. The image quality at the position of ~8.5 cm (Fig. 5(d)) is still high enough to distinguish the multilayer structures of the tape without much degradation of resolution and sensitivity. It is observed that the image brightness of the multi-layered tape is not uniform, where darker bands appear below the surface of the tape. The reason for this non-uniform brightness is likely due to the polarization artifact introduced by the birefringent property of the tape. Nevertheless, this experiment demonstrates well that the high quality images can be maintained over a ~10 cm ranging distance by using our UW-OCT with the proposed calibration procedure, showing the feasibility of imaging large-volume samples where the long ranging distance is required for practical imaging.

Fig. 6. A front view of the 3D OCT image rendering of human hand captured within ~2.5 sec. The size of the image is 12 cm (vertical) x 20 cm (horizontal).
3.2 OCT imaging of whole human hands in vivo

After testing the feasibility of the UW-OCT capable of the ultra-long ranging distance, we further conducted in vivo OCT imaging of human hands and faces, which is more attractive for future applications. The preliminary study that uses home-built systems to image human subjects was reviewed and approved by Institutional Review Board of University of Washington, and the informed consent was obtained from all subjects before imaging. This study followed the tenets of the Declaration of Helsinki and was conducted in compliance with the Health Insurance Portability and Accountability Act.

Figure 6 shows a front view of 3D OCT image captured from the right hand of an adult male volunteer. By scanning the probe beam through the wide-angle camera lens in the sample arm, the FOV of ~12 × 20 cm² was designed to cover the entire right hand. In order to avoid the overflow of high data throughput from the on-board memory of the digitizer to the host computer, we captured 500 A lines per B frame in the fast scanning axis and 500 locations in the slow scanning axis with the sampling rate of 3.2 GS/s for 3D OCT imaging. This procedure generated a large data of 15 GB per volume. Due to the swept rate of the akinetic source at 100 kHz, the data acquisition time was accomplished within ~2.5 second. In Fig. 6, all the fingers as well as a large area of palm and a small part of wrist are contained within the FOV of a single OCT imaging, which has never been demonstrated before. The main feature of the palm print and the profile of phalanges in the fingers are clearly identified.
Figure 7(a) is the corresponding cross-sectional image at the location marked by the red dashed line in Fig. 6, where we observe all of the five fingers simultaneously in a single B scan. Figure 7(b) is the cross-sectional image at the location of the palm as indicated by the yellow dashed line in Fig. 6. The cross-sectional structure over a large area is expected to provide useful information to quickly evaluate or identify the health condition of human skin. The information content is rich because of the 3D imaging capability of OCT. Figures 7(c) and 7(d) are the further zoom-in images with dense sampling of the areas as indicated in the green box in Fig. 7(a) and the blue box in Fig. 7(b), respectively. Because the axial resolution of the system is ~13 microns in air, the zoom-in image of the middle finger enables us to identify epidermis, dermis and small blood vessels in the superficial layer of the skin as shown in Fig. 7(c) where one can appreciate that the image quality is similar to that obtained from conventional short range OCT systems. In the zoom-in image of the palm, Fig. 7(d), we are able to visualize a large blood vessel in the deep tissue. Please note that the longer ranging distance does not mean the improved penetration depth for OCT imaging. The imaging depth is however limited by the optical properties of the tissue target. Here, because human skin is of high scattering properties, the maximal depth is typically less than 2 mm as shown.

![Figure 7(a) and 7(b)](image)

Fig. 8. OCT portraits of (a) a glass wearing young man with neutral expression and (b) another young man with mouth open. The size of the portraits is 21 cm (vertical) x 18 cm (horizontal).

### 3.3 OCT imaging of whole human faces in vivo in one scan

After the successful demonstration of the entire human hand imaging, we further performed the large-volume OCT imaging for human faces to demonstrate the advantages of our developed UW-OCT system. During the imaging, the volunteer laid down with the face up and with the head positioned on a cushion pillow, while the OCT probe was mounted on an articulating arm with the probing light pointing down for ease of imaging. For this experiment, the imaging FOV was further expanded to ~21 × 18 cm² to cover the whole area of human face. Figures 8(a) and 8(b) show the front view of the 3D rendering of OCT data sets captured from two young volunteers with different facial expressions. With the focus of the probe beam positioned approximately at the middle depth position of human face, the neutral facial expression (Fig. 8(a)) with the structures including the forehead, eyebrows, eyes, ears, cheeks, nose, mouth and chin are clearly seen, which may be very useful in providing important information for human face recognition and communication. Due to the ultra-long coherence length in our UW-OCT system, the pillow used to comfort the volunteer for lying down is also visualized at the deep position under the head. In addition, the high-speed akinetic swept laser with fast data acquisition enabled us to take a snapshot of the facial
expressions. A close-up shot of a young man face with mouth open was demonstrated in Fig. 8(b). The results demonstrate amply the capability of our developed UW-OCT system for making the human face 3D portraits from just a snapshot of a few seconds, which we believe would bring new opportunities for future OCT imaging applications.

Fig. 9. (a) The cross-sectional image at the position of human eye as marked by the green dashed line in Fig. 8(a). (b) The cross-sectional image at the position of human mouth as marked by blue dashed line in Fig. 8(b).

Figure 9(a) provides the cross-sectional OCT structure at the position of eye (marked by the green dashed line in Fig. 8(a)). In this figure, we are able to visualize the top and bottom surfaces of the eye glasses and profile of the nose. With the light penetrating through the thick glass wear, the skin around the eye and the eyelash are also well identified. Moreover, the biometric structural information of human anterior segment of eye, including entire transverse anterior chamber width, axial eye length, anterior chamber angle, iris and entire crystalline lens, are clearly observed within this single cross-section image. Such rich biometric information useful for ophthalmological applications is enabled by the capability of long depth range and wide FOV of our UW-OCT system. Figure 9(b) illustrates a representative cross-sectional image at the position of mouth as indicated by the blue dashed line in Fig. 8(b). With the mouth open in the figure, the light can penetrate into the teeth to provide an ability to resolve the internal structure from top to bottom in the OCT cross-sectional image,
demonstrating that our UW-OCT system maybe valuable for the new applications in the diagnosis and evaluation in dentistry.

4. Discussion and conclusion

In this study, we demonstrated a state-of-the-art UW-OCT system with ultra-long ranging distance and ultra-wide FOV by taking the advantages of stable and programmable Insight laser source that can be digitally swept without any mechanical moving parts. The capability of ultra-long imaging range was resulted from the single-mode operation of the swept source, giving a long conference length (>10 cm), and the heavy oversampling of spectral interferograms through data acquisition. Combining the inherently linear sweep behavior at the factory setting together with the proposed adaptive calibration procedure for the extended depth, the akinetic swept laser provided unprecedented roll-off performance without a need for external k-clocks. Benefiting from the highly repeatable and stable akinetic swept laser, the durability of the adaptive calibration is equivalent to the durability of the laser’s self-calibration. In our studies, a new calibration is only necessary when the laser sweep is re-programmed/re-calibrated.

It has been previously reported that the akinetic laser employed in this study was capable of operating at different sweeping rates ranging from 20 kHz up to 200 kHz through programming control digitally [21]. The flexible tunability of the sweeping rate promises to provide scalable sensitivity in functional OCT without data redundancy. In this study, we employed an akinetic swept source of 100 KHz to achieve OCT imaging with ultra-wide FOV (~20 x 20 cm²) and ultra-long ranging distance (~10 cm) that required an imaging time of ~2.5 sec. To improve this imaging time, the sweeping speed of the laser source clearly needs improved. It is believed that the A-scan rate of the akinetic laser can be further developed to operate at megahertz range in future that would meet the requirement of faster imaging rate. In our system, the spacing between adjacent A-scans was ~190 µm for an FOV of ~20 cm x 20 cm. The faster sweeping rate of the laser source would be useful to improve this spacing to deliver high definition imaging of the large volume samples if needed. On the other hand, the faster sweeping rate of the source would be also more favorable for the future development of high-speed optical coherence elastography [23], 4D angiography imaging [24] and Doppler OCT to measure the velocity of the blood flows.

While promising, future developments to increase the volumetric imaging rate while maintaining the demonstrated merits of the current system require a number of challenging advancements in the aspects of light source, data acquisition and computing power. Provided successful developments, even more challenging is the issue in the optimal implementation of these advancements in an OCT imaging system because these aspects are unfortunately coupled. For example, the improvement of the OCT imaging rate requires the increase of the sweeping rate of the laser source, which in turn demands faster data acquisition to maintain the ranging distance, e.g. at more than 10 cm. Faster data acquisition, on the other hand, demands the more advanced computing system with sufficient interface bandwidth to handle the large volume of data flowing from the OCT system into the computing device. It is hopeful that with the current momentum of high-speed data acquisition, high-throughput data transition and super-power computation (e.g. GPU processor), the big data pressure for large-volume OCT imaging could be relaxed.

In this study, the actual sensitivity roll-off in 3-dimentional OCT imaging is also affected by the sample arm optics. The de-focusing effect of the object lens is another major factor that limits the depth of focus, while the object is outside the depth of focus of the objective lens, the sensitivity is significantly reduced. However, the limitation of OCT beam delivery is not the focus of this work, the optimal beam delivery method for ultra-long depth range OCT still remains to be explored.

Given the feasibility demonstration of the UW-OCT imaging to provide ~10 cm ranging distance and ~400 cm² FOV, there is a number of new opportunities that could be envisioned
for future OCT applications. Below are some promising examples: 1) For biomedical imaging applications, the UW-OCT system would be particularly useful to serve the purpose of a wide field of view inspection of the tissue states to quickly identify the loci of the tissue injury or lesions for further detailed clinical investigation, for example in the case of skin burns where the area of the burn is often large and the skin surface is typically uneven with surface elevation sometimes more than centimeters. 2) Due to its high resolution and 3D imaging capabilities, the UW-OCT system may be useful in precisely determining the interpupillary distance (IPD), defined as the distance between the center of the pupils of the two eyes (e.g., Fig. 8(a)). IPD is critical for the design of binocular viewing systems, where both eye pupils need to be positioned within the exit pupils of the viewing system. These viewing systems include binocular microscopes, night vision devices or goggles, and head-mounted displays. The precise measurement of the IPD data among individuals (which is 3D in nature) are important in the design of such systems to specify the range of lateral adjustment of the exit optics or eyepieces. The precise measurement of the IPD is also useful to specify prescription eyewear in optometry. 3) In security, the 3D imaging with depth-resolved information is capable of providing a unique biometric identification (ID), which promises to identify spoofing attacks for high-security applications because OCT can provide the structural information beneath surface, which however is impossible for the current picturing approach using digital cameras. The OCT portrait of human faces could also be utilized in facial recognition that serves for security purposes. Using the human face as a key to security, biometric face recognition using 3D OCT portraits may have its potential for a wide variety of applications in both law enforcement and non-law enforcement. Here, the 3D face images can be captured by an OCT system from a distance without touching the person being identified. In addition, such 3D face recognition could serve the crime deterrent purpose because face images that have been recorded and archived can later help identify a person. 4) The UW-OCT would have potential to provide precise localization information for 3D printing. And 5) it promises to serve a wide variety of non-biomedical applications, such as non-destructive inspection of smart phone and quality control of produces, e.g. tomato or potato, in agriculture industry.

In summary, we have demonstrated an ultra-long range and ultra-wide OCT system for large-volume imaging based on an akinetic swept source. We have proposed a simple strategy of adaptive calibration to further optimize the imaging performance in the extended depth range beyond the factory default setting of sweeping rate in the laser source, delivering unprecedented imaging performance for our UW-OCT system. Utilizing an ultra-fast digitizer with wide field optical and mechanical components, we have achieved up to ~10 cm long ranging distance with as large as up to 20 × 20 cm² FOV. This system enabled us to succeed in the demonstration of imaging entire human hands and faces at one single scan session of ~2.5 sec. It is noteworthy that, to the best of our knowledge, this is the first demonstration of in vivo OCT imaging for such a large volume biological sample in a single acquisition. The advancements we have achieved in this study are expected to open up new opportunities for future imaging applications of OCT in a variety of fields, including both biomedical and non-biomedical applications.

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