Scalable multiplexing for parallel imaging with interleaved optical coherence tomography

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Abstract: We demonstrate highly parallel imaging with interleaved optical coherence tomography (iOCT) using an in-house-fabricated, air-spaced virtually-imaged phased array (VIPA). The air-spaced VIPA performs spectral encoding of the interferograms from multiple lateral points within a single sweep of the source and allows us to tune and balance several imaging parameters: number of multiplexed points, ranging depth, and sensitivity. In addition to a thorough discussion of the parameters and operating principles of the VIPA, we experimentally demonstrate the effect of different VIPA designs on the multiplexing potential of iOCT. Using a 200-kHz light source, we achieve an effective A-scan rate of 3.2-MHz by multiplexing 16 lateral points onto a single wavelength sweep. The improved sensitivity of this system is demonstrated for 3D imaging of biological samples such as a human finger and a fruit fly.

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Introduction

The ability to perform high-speed imaging has played an important role in growing the acceptance of the clinical utility of optical coherence tomography (OCT) [1]. In particular, increasing the speed of OCT systems can improve clinical diagnoses, because faster systems are less susceptible to motion artifacts, can image larger areas, and can produce higher quality data (i.e., better SNR through averaging) [2]. High-quality imaging of biological samples is often associated with sensitivities on the order of ~90dB or higher. Because sensitivity is proportional to input power, however, the light exposure limitations imposed by national safety standards lead to a well-known trade-off between imaging speed and sensitivity in the raster-scanned system designs that are standard for most OCT implementations (i.e., where one lateral position is imaged at a time). As others have recognized, one way to overcome the speed/sensitivity tradeoff is to implement parallel imaging schemes in which many lateral positions are imaged simultaneously. These schemes work because the power threshold for safe light exposure scales with imaging area; thus, increasing the number of lateral positions that are imaged allows one to image more points in less time while increasing the power to maintain high sensitivity.

Parallel imaging schemes have been demonstrated for both spectral-domain (SD-) and swept-source (SS-) OCT systems. Examples of the former include streak-mode detection [3], parallel Fourier-domain OCT [4,5], multi-photoreceiver array methods [6], and spectrally encoded endoscopy [7,8]. In the case of the latter, several multi-point imaging schemes for SS-OCT have been introduced that multiplex the acquisition of different lateral positions in time [9,10] or space [11–14]. Unfortunately, such schemes are not easily scalable for multiplexing many lateral points because the components needed to perform the multiplexing are expensive and require ample physical space (e.g., duplicated optics, additional detectors,
or custom fiber splitters with long fiber spools to create optical delay lines); moreover, such multi-point schemes are not capable of eliminating the need for scanning elements to create contiguous images because of the finite separation between imaged points. This is important because the mechanical limitations of scanners can ultimately become a bottleneck for imaging speed [15].

In contrast to the multi-point parallel schemes demonstrated for SS-OCT, we recently introduced interleaved-OCT (iOCT), a line-imaging scheme that uses a virtually imaged phase array (VIPA) to perform multiplexing of multiple, contiguous lateral points in the spectral domain [16]. This line-imaging scheme is, in principle, scalable to multiplex many lateral points (> 100) if using a light source with sufficiently high power. The iOCT concept demonstrates that long coherence lengths can be exploited to perform parallel imaging by repurposing excess ranging depth to image different lateral points.

Our previous work demonstrated the versatility of the iOCT concept by showing that the design is wavelength-independent and that the same system design could be used with sources operating at 10-, 20- and 100-kHz sweep rates. We performed iOCT with a custom-designed glass VIPA and demonstrated the ability both to acquire biological images and to capture data with effective A-scan rates of up to 1 MHz using a 100-kHz source. However, the cost, fabrication time and rigid nature of the commercially fabricated glass VIPA left little flexibility in the design of the iOCT system: the number of multiplexed points was fixed. Moreover, the sources used in that demonstration were not ideal for iOCT because of their inherently low source power and short coherence lengths, which limited the quality of the images and the achievable ranging depths.

In this work, we introduce a new design for iOCT that also incorporates an in-house-fabricated, air-spaced VIPA with tunable parameters. Compared to our former system, this new system boasts higher sensitivity, a faster base speed (200-kHz sweep rate), and a longer coherence length (which leads to a longer ranging depth) due to the use of a 1310-nm VCSEL [17,18]. Additionally, the use of an air-spaced VIPA that we fabricated ourselves allows us to tune the number of multiplexed points as desired for a given imaging application to balance the need for speed, ranging depth and good sensitivity, making it truly scalable. This flexibility in the design of the system is a clear advantage of iOCT over other parallel imaging techniques that have been demonstrated that require extensive modifications to system hardware. The air-spaced VIPA we use in this work is simple enough to be fabricated by undergraduate students in a standard cleanroom and is ten-times cheaper than the commercial glass VIPA used in our previous work.

To better understand the effect of various parameters of the VIPA on the performance of iOCT, as well as practical considerations for constructing iOCT systems, herein we also provide a thorough discussion of the parameters and operating principles of the VIPA as relate to parameters of interest for OCT imaging.

2. Backgrounds and theory

2.1 General overview of iOCT

Simply stated, iOCT sends a sequence of unique frequency combs to different lateral positions in the imaging plane; all combs are generated simultaneously through the use of a multi-beam demultiplexer (MBDX) and have the same frequency spacing (given by the free spectral range) but comprise different sets of wavelengths. At the detector, all combs are detected simultaneously, and a simple demultiplexing step is performed in the software to extract the individual comb-like interferogram associated with each lateral position from the multiplexed interferogram sampled by the detector. A more complete discussion of the processing steps involved in reconstructing A-scans is found in prior work [16].
2.2 VIPA characteristics and OCT parameters of interest

While there are many ways to implement the functionality of the MBDX, a VIPA [19] provides one of the simplest and most robust designs for converting a standard OCT system into an iOCT system (see Fig. 1(a)). A VIPA is essentially a cross between a Fabry-Perot etalon and a Gires-Tournois interferometer: it consists of two surfaces with reflectivities $R$ and $r$ and its finesse is given by Eq. (1). $R$ is typically a high-reflectivity surface ($R \approx 1$).

$$\text{finesse} = \frac{\pi \sqrt{Rr}}{1 - Rr} \quad (1)$$

A VIPA differs from a standard Fabry-Perot etalon in that it contains a small entrance window (usually opened, uncoated or anti-reflective-coated) into which line-focused light is coupled via a cylindrical lens. All wavelengths enter the VIPA at a range of incident angles given by the numerical aperture of the cylindrical lens and propagate in the VIPA according to the Law of Reflection; the exact angles of reflection are determined by the angle $\theta_i$ at which the VIPA is tilted relative to the optic axis of the OCT sample arm (see Fig. 1(c)). As with a standard etalon, light accumulates phase while reflecting inside the cavity due to the optical pathlength traversed. Light exiting the etalon is made to interfere using a standard spherical lens (SL1) placed at the output of the VIPA, and the condition of constructive interference is met only at discrete angles of reflection. The angle corresponding to constructive interference is unique to each wavelength and is associated with a unique angle of refraction upon exiting the VIPA; the presence of spherical lens SL1 causes light from different angles to illuminate different lateral positions in the image plane, as in a standard telecentric imaging system.

The free spectral range (FSR) of the VIPA is defined as the minimum spacing between two wavelengths for which the angle of constructive interference is the same; such wavelengths are part of the same frequency comb (albeit from different diffraction orders) and map to the same lateral position in the imaging plane. Hence, all of the wavelengths that
are transmitted to a particular lateral position are equally spaced by the FSR. The FSR (in units of meters) is a function of a center wavelength of the source $\lambda_C$, the refractive index $n$, the thickness of the VIPA $t$, and the tilt angle of the VIPA relative to the optic axis of the system $\theta_i$ (Eq. (2)) [20,21]:

$$FSR_A = \frac{\lambda_C^2}{2t \cdot \cos(\theta_i) - 2t \cdot \tan(\theta_i) \cos(\theta_i) \theta_i - t \cdot \cos(\theta_i) \theta_i^2 / n}.$$  

(2)

Here $\theta_i$ is the internal angle inside the VIPA, which differs from $\theta_i$ only in the case of a dielectric-filled cavity, and is determined by Snell’s law: $n \cdot \sin(\theta_i) = \sin(\theta)$. The goal of iOCT is to acquire $M$ interferograms from $M$ lateral positions during a single sweep of the laser source. iOCT thus relies on high-bandwidth photodetectors and high-speed digitizers to sample many more points than one does in a traditional OCT system (ideally $M$-times as many). The maximum possible number of uniquely sampled points across the entire bandwidth of the source, each assumed to correspond to a different wavelength, can be limited by the coherence length of the source (or alternatively, its instantaneous linewidth), the sampling rate of the digitizer, or the bandwidth of the photodetector. We use the term “resolvable points” ($RP$) to refer to the maximum number of unique wavelength samples that can be captured by the system; this number sets an upper limit on the multiplexing capabilities of iOCT. Ultimately, the maximum value of $M$ is determined by the relation $RP = N \times M$, where $N$ is the number of points used to sample the interferogram for a single lateral position (i.e., the number of points in the comb); hence, for a fixed value of $RP$ there is an inherent trade-off between $N$ and $M$, the number of points per interferogram and the number of multiplexed A-scans per iOCT B-scan, respectively.

Any optical element in the system that can induce spectral cross-talk, including the VIPA, can reduce the effective value of $RP$. Note that while cross-talk in many depth-multiplexing schemes is manifested as a degradation in the image due to overlapping image structures, cross-talk in iOCT results in a degradation in lateral resolution. To understand how various aspects of the design of the system and the VIPA affect parameters of interest to OCT, we consider the case where $RP$ is limited by the VIPA and not by other components in the system. Table 1 provides a list of important parameters in the design of an OCT system, the formulas that govern them, and the parameters and components of the iOCT system that affect their values. Three sets of parameters characterize the operation of the VIPA. The first set depends on the VIPA itself: its thickness ($t$), the reflectivity of the entrance ($R$) and output ($r$) dielectric interfaces, the index of refraction of the VIPA material ($n$) and its tilting angle with respect to the propagating direction of the light ($\theta_i$). The second set of parameters depends on the light source; specifically, its center wavelength $\lambda_C$, instantaneous linewidth $d\lambda$, and full bandwidth $\Delta\lambda_{Total}$. The third set of parameters depends on the optical components in the system: that is, the focal length of the spherical lens ($f_{SL1}$) and the magnification factor of the 4-f system.

As in standard OCT, the axial resolution of iOCT is determined by the center wavelength and bandwidth of the source. This value is not affected by the VIPA or other optical components (except to the extent that the coatings of the components limit the wavelength passband).

As in standard OCT, the ranging depth of each A-scan, which is the deepest position of the sample that can be imaged without aliasing according to the Nyquist limit, is a function of the sampling frequency. In iOCT, the sampling frequency is equivalent to the spacing of the frequency comb at each lateral position, which is determined by the FSR. The FSR is most directly a function of the optical thickness of the VIPA; for the air-spaced VIPA used in this work, the FSR can be adjusted by changing the thickness of the VIPA – that is, the gap distance between the two reflecting surfaces.
### Table 1. List of important parameters in the design of an OCT system

| iOCT parameters                  | Formula                                      | Source parameters                  | VIPA parameters       | System parameters  |
|----------------------------------|----------------------------------------------|------------------------------------|-----------------------|---------------------|
| Axial resolution                 | \( \frac{2 \ln 2}{\pi} \frac{\lambda^2}{\Delta f_{\text{FWHM}}} \) | FWHM bandwidth*, Center wavelength | VIPA parameters       | System parameters  |
| Ranging depth                    | \( \frac{1}{4} \frac{\lambda^2}{\Delta \lambda_{\text{FSR}}} \) | Center wavelength*                 | FSR                   |                     |
| # points / A-scan               | \( \Delta \lambda_{\text{Total}} / \Delta \lambda_{\text{FSR}} \) | Total bandwidth*                   | FSR                   |                     |
| # resolvable lateral points     | \( \frac{\pi \sqrt{R}}{1 - R} \)             | Linewidth                          | *finesse*             |                     |
| Lateral imaging range           | \( \frac{\lambda_{\text{c}}}{2 \theta \cdot t} \) | Center wavelength*                 | Thickness, incidence angle, *finesse*, *system magnification* |
| Lateral resolution              | \( \frac{\lambda_{\text{c}}}{2 \theta \cdot t \cdot *finesse*} \) | Center wavelength*                 | *finesse*, *incident angle*, thickness*, *finesse*, *system magnification* |
| Sensitivity                     |                                               |                                    | FSR, *finesse*        |                     |

* indicates that the corresponding iOCT parameter is proportional to this parameter.

As with standard OCT, the number of points per A-scan is a function of the sampling frequency of the interferogram for an individual lateral position and the total bandwidth of the source. Because the sampling frequency with iOCT is simply the FSR, the number of points per A-scan \( N \) is determined by the ratio of the total bandwidth of the source to the FSR. Note that it is important that the digitizer be fast enough to achieve the desired sampling rate. As with standard OCT, the number of points per A-scan in iOCT can be increased by increasing the total bandwidth of the source.

For a given FSR, the number of resolvable lateral points that can be multiplexed, or equivalently the number of resolvable points per iOCT B-scan, is determined by the number of resolvable points that can be sampled within the FSR. The larger of the *finesse* of the VIPA and the instantaneous linewidth of the source characterizes the narrowness of the wavelength samples that can fit within the FSR. Thus, increasing the number of multiplexed points \( M \) can be accomplished by increasing the *finesse* and using a source with a narrow linewidth. Note that the *finesse* is a function of the reflectivity of the VIPA surfaces; high *finesse* requires high reflectivity.

The lateral image range, or the length of the line illuminating the image plane, is controlled by both the VIPA and the optical components. The two distal ends of the line beam, which arise from the two largest output angles of the VIPA, span the smallest and the largest wavelengths within a FSR. The wavelength span of these angles, which is related to the spectral dispersion of the VIPA, is governed by the following equation [20]:

\[
2t \cos \theta_m - 2t \tan \theta_m \cos \theta_i \cdot \theta_{\lambda, m} + t \tan \theta_m \sin \theta_i \cdot \theta_{\lambda, m}^2 = m \lambda. \tag{3}
\]

Once the thickness (or FSR) is set, each wavelength will constructively interfere at multiple angles \( \theta_{\lambda, m} \), associated with different diffraction orders, \( m \). Because spectral encoding with iOCT assumes a one-to-one mapping between wavelength and spatial position, it is important to avoid such spectral cross-talk. In practice, this means that the angular range of spectral dispersion should be limited to the difference between consecutive diffraction orders, \( m \cdot \lambda \) and \( (m + 1) \cdot \lambda \); thus, \( \Delta \theta_{\lambda, m} = \frac{\lambda}{(2 \theta_i \cdot t)} \), which is proportional to the lateral line length. Typically \( \theta_i \) should be as small as possible to guarantee that light coupled into the cavity has the maximal spatial overlap to achieve effective interference. However, effective imaging with
small values of $\theta_i$ requires large values of $t$ because the size of the entrance window decreases with decreasing $t$. Therefore, the product $\theta_i \cdot t$ is nearly fixed for all possible values of $t$, resulting in a nearly fixed value for the lateral range.

The lateral image range further depends on the focal length of the lens placed after the VIPA (SL1). Additional optics can be placed in the system to magnify or demagnify the length of this line when relaying it to the sample plane (e.g., with a 4-f optical system after the VIPA). One should take caution, however, when increasing the length of the lateral line prior to the galvanometer to ensure the beam is not clipped by its aperture or by any downstream components (e.g., additional lenses or galvanometers).

The lateral resolution of the iOCT system is simply the lateral image range divided by the finesse. As with standard OCT, the diffraction limit serves as a lower bound on this value.

The sensitivity of the iOCT system is mainly affected by the insertion loss of the VIPA. The loss is primarily caused by the effect of the surface with reflectivity $r$ on the back-coupling efficiency of the VIPA. The higher the value of $r$, the higher the finesse (Eq. (1)) but the lower the back-coupling efficiency and the lower the sensitivity. Because the finesse determines $M$, there is an inherent tradeoff between $M$ and the sensitivity: that is, iOCT is most efficient for low to moderate values of $M$. In addition, the FSR determines the number of points per A-scan ($N$), which is proportional to the sensitivity since, as in standard OCT, the noise floor decreases with more samples [22].

Based on Eq. (1)-(3), we simulated the spectral output of an air-spaced VIPA for different design conditions. The expected output angles associated with various wavelengths were calculated and are plotted in Fig. 2. The baseline design condition was set as $t = 2.5$ mm, $r = 50\%$, and $\theta_i = 3.2^\circ$ as shown in Fig. 2(b). The finesse was ~5 and the FSR was 344pm. When the thickness was doubled to $t = 5.0$ mm and the incidence angle was correspondingly halved to $\theta_i = 1.6^\circ$, the FSR was halved to 172pm, and the finesse remained unchanged (see Fig. 2(a)). When the reflectivity $r$ was increased to 80\% compared to the baseline condition, the finesse tripled to ~15 while the FSR was unchanged, as shown in Fig. 2(c). These simulation parameters were chosen to match the experimental conditions described below.

Fig. 2. Simulation of iOCT showing expected output angle as a function of wavelength for an air-spaced VIPA with three different conditions. The baseline setting is shown in (b): $t = 2.5$ mm, $r = 50\%$, and $\theta_i = 3.2^\circ$. (a) Lowering the FSR (by increasing the thickness) increases the sampling frequency of the interferogram at a given lateral position, which increases the ranging depth. (c) Increasing the finesse by increasing the reflectivity translates into a higher multiplexing factor (and higher effective A-scan rate).

3. Methods

3.1 Fabrication of the VIPA

The finesse of the VIPA can be controlled by changing the reflectivity of its surfaces. To test the effect of finesse on the performance of iOCT, we fabricated two VIPAs using in-house clean-room facilities. The first surface of the VIPA consisted of a commercially purchased D-shaped mirror with an HR coating ($R \approx 100\%$). The second surface was made by e-beam evaporation of sequentially deposited layers of Cr and Au (35:35 Å and 100:100 Å for the
two VIPAs, respectively) onto a 1-inch-diameter, polished optical flat. To achieve the target layer thicknesses, the following evaporation conditions were used: base pressure, $7 \times 10^{-7}$ Torr; deposition rate, $\sim 0.5-1$ Å/s; source-to-substrate distance, 26.8 cm; material purity, 99.999%; electron beam energy, 1 kV. The emission current was varied as necessary to achieve the desired deposition rate. The resulting reflectivities of the second VIPA surface were $r = 45\%$ and $r = 80\%$, respectively.

3.2 Data processing and calibration

In this work, we used a VCSEL swept source (Thorlabs Inc.) that has a high sweep rate (200-kHz) and large coherence length (>50 mm). The power out from the source was 34mW. The balanced photodetector we used had an electronic bandwidth of 800 MHz (1.6 GHz, PDB480C, Thorlabs Inc.) and signals were digitized (ATS9360, AlazarTech) at a sampling rate of 1.2 GSa/s, which was primarily limited by the bus rate of the board for streaming the data (3.6 GB/s, PCIE Gen 2). The phase of the OCT signal was stabilized by referencing it to a Mach-Zender interferometer (MZI) signal collected from the second channel of the digitizer.

All data were processed using customized Labview software. Low-pass filtered versions of the collected interferograms were subtracted from their raw counterparts to remove DC and fixed-pattern noise. Wavelength data were interpolated into linearly spaced wavenumbers based on the unwrapped phase of the Hilbert-transformed MZI signal. The spectral envelope for the interferogram was reshaped with a Hanning window in order to reduce the sidelobes. For each sweep, we digitized 4,096 points; however, interferograms were interpolated to contain 8,192 data points in total to minimize any errors caused by resampling [23]. As described previously [16], processed interferograms were down-sampled using an array of indices corresponding to the positions of the wavenumbers associated with a given lateral location. The indices were decided in advance based on a pre-calibration step. Each demultiplexed interferogram was zero-padded prior to taking the FFT to reconstruct the A-scan for that lateral position. Dispersion compensation was performed in hardware by matching the optical paths in the reference and sample arm by using additional optics.

4. Results and discussion

4.1 Experimental test of iOCT with an air-spaced VIPA

We monitored the effect of different design parameters for the air-spaced VIPA on the spectral output of a single lateral position on a reflective sample (mirror): we used a mechanical slit to restrict the reflected light to a small portion of the mirror and blocked the reference arm. Measurements were taken at two different lateral locations (black solid vs. gray dotted graph); the peaks represent the wavelengths associated with the interrogated lateral point. As in Fig. 2, consider the result in Fig. 3(b) as the initial condition. Decreasing the FSR by increasing the thickness of the VIPA (see Fig. 3(b) → Fig. 3(a)) reduces the distance between the peaks. This is equivalent to increasing the sampling frequency in the $k$-domain, which will improve the ranging depth of the OCT system. The sensitivity is also improved because there are more sampled points in the interferogram. It is noticeable that the finesse is also reduced, albeit slightly, which could be due to imperfections in the system alignment after the change. Increasing the finesse relative to the initial condition by using a VIPA with a higher reflectivity (see Fig. 3(b) → Fig. 3(c)) improved the sharpness of the peaks (and lateral resolution) but leaves the peak separation unchanged (i.e., the FSR is unchanged). This is equivalent to improving the multiplexing capability and lateral resolution of the system without changing the ranging depth; however, high reflectivity will decrease the sensitivity of the system. The specifications of the system associated with each case are listed in Table 2.
Fig. 3. Spectral output from a single lateral position of the sample for three VIPA conditions showing experimental verification of the simulations in Fig. 2. The parameters of the VIPAs are given in Table 2. The FSR, FWHM peak width of each peak, and the finesse are calculated and shown in each graph.

Table 2. Specifications of the three VIPA systems

|                | Type 1: Low finesse, Low FSR | Type 2: Low finesse, High FSR | Type 3: High finesse, High FSR |
|----------------|-------------------------------|-------------------------------|-------------------------------|
| Reflectivity ($r$) | 45%                           | 45%                           | 80%                           |
| Thickness ($t$)    | 5.0 mm                        | 2.5 mm                        | 2.5 mm                        |
| Incidence angle ($\theta$) | 1.6°                          | 3.2°                          | 3.2°                          |
| Finesse           | 3.4                           | 4.4                           | 7.1                           |
| FSR              | 162 pm                        | 326 pm                        | 325 pm                        |
| Axial resolution  | 12.2 $\mu$m                   | 12.2 $\mu$m                   | 12.0 $\mu$m                   |
| Lateral resolution| 19.7 $\mu$m                   | 15.6 $\mu$m                   | 12.4 $\mu$m                   |
| Lateral range     | 65 $\mu$m                     | 65 $\mu$m                     | 65 $\mu$m                     |
| iOCT Sensitivity  | 93.1 dB                       | 89.1 dB                       | 84.2 dB                       |

4.2 Demonstration of 10 × multiplexing with a large ranging depth (M = 10)

To demonstrate the ability of iOCT to perform biological imaging, volumetric data sets of a human finger were collected using the air-spaced VIPA with a low finesse and low FSR (Type 1 in the Table 2, $r = 45\%$, $t = 5$ mm). The power on the sample was 7 mW. The volume consists of 10 B-scans comprising 5,000 A-scans in total: 10 lateral points were multiplexed onto each sweep and the beam was scanned in 1-D during 500 sweeps to create 10 B-scans simultaneously. Figure 4 shows the first and the last B-scans collected during the sweep. The large ranging depth (2.6 mm) enables visualization of all subsurface structures throughout the entire lateral scanning range (8.0 mm). The insets highlight the appearance of sweat ducts that appear clearly in the first B-scan but are not present in the last B-scan. The first and last B-scans are separated by the length of the lateral image range (65 $\mu$m), which is approximately the size of a sweat duct [24]. Note that the ghost images that appeared when a glass-VIPA was used [16] could be effectively removed by properly setting the thickness of the air gap of the air-spaced VIPA such that all the reflection images are overlapped. The sensitivity of the system was 93.1 dB for this configuration.

A larger volumetric data set of a human finger was then collected by using a second galvanometer to perform additional scans at several lateral locations orthogonal to the direction of the B-scan. The data set shown in Fig. 5 consists of 50 sub-volumes, each of which contains 3,000 A-scans (10 × 300): 10 multiplexed lateral points with 300 A-scans / B-scan (effective A-scan rate = 2 MHz). In total, the volume size was 612 (depth) × 300 (fast scan axis) × 500 (slow scan axis), corresponding to a volumetric scan area of 2.6 mm × 3.2 mm × 3.2 mm. The volume image is rendered in 3-D as shown in Fig. 5(a). Figure 5(b) shows two ortho-slices where some sweat ducts are clearly visible.
Fig. 4. Simultaneously acquired B-scan images of a human finger obtained with iOCT: 8.0-mm wide x 2.6-mm deep using an air-spaced VIPA with low finesse and low FSR. The volume consists of 5,000 A-scans (10 × 500) with 10 lateral points multiplexed in each sweep and 500 sweeps with the beam scanned in 1-D. The first (a) and the last (b) slices are separated by 65 μm. (Scale bar: 500 μm x 500 μm)

Fig. 5. Volumetric images of a human finger. Volume rendered view (a) and two orthogonal slices from the volume set (b). The volume consists of 612 × 300 × 500 points along the depth, fast scan axis, and slow scan axis, respectively. The scan area was 3.2 mm × 3.2 mm and the ranging depth was 2.6 mm.

4.3 Demonstration of 16 × multiplexing with smaller ranging depth (M = 16)

To show the scalability of the multiplexing factor using an air-spaced VIPA, we took images of fruit flies using an air-spaced VIPA with higher FSR (Type 2 in the Table 2, $r = 45\%$, $t = 2.5$ mm). The ranging depth was 1.3 mm and the number of multiplexed positions was increased to 16, resulting in a 3.2-MHz effective A-scan rate. The power on the sample was 7 mW, and the sensitivity of this system was 89.1 dB. Figure 6 shows a representative volume image of a fruit fly. A video fly-through of the data is also included as a supplement to the main text.
The number of multiplexed points is currently limited by the VIPA *finesse*, the source linewidth, and the digitizer speed. One advantage of using a VIPA as the MBDX for iOCT is that the *finesse* can be easily be made to exceed 1000 by increasing the reflectivity; moreover, source linewidths and digitizer speeds have been steadily improving in recent years [17]. Hence, it is conceivable that iOCT can eventually be used to perform fully scan-less imaging – that is, obtaining a B-scan image comprising > 500 A-scans from a single laser sweep without any beam scanning. This strategy could be particularly ideal for endoscopic OCT, where the size of the scanning element limits the size of the endoscope. Fabry-Perot interferometers with comparable designs to the VIPA can be easily miniaturized with micromachining [25–27]. Scan-free imaging is also an important consideration for high-speed imaging at effective A-scan rates exceeding a few MHz, at which point the volumetric imaging speed is actually limited by the scanning rate [15].

The sensitivity of our current system has been improved compared to the previous work by utilizing a VIPA with lower reflectivity, and by improving the source and optical system design. However, a target sensitivity of higher than 100 dB would be desirable for clinical applications. The current sensitivity is partly limited by our source power: a sensitivity of 104.5 dB would be obtainable if we were operating at the maximum permissible exposure (MPE) limit allowed by ANSI standard for continuous illumination. When operating at the MPE, the obtainable sensitivity is equivalent to that using traditional SS-OCT with an ultrafast swept source having a sweep rate equivalent to the effective A-scan rate of iOCT (with multiplexing), because the lengths of practically achievable lateral lines does not exceed the size of the limiting aperture for ANSI standards. As has been noted by others, the MPE – and therefore the sensitivity – could be further increased in the case of a pulsed illumination protocol in which the scanning and the illumination end once the data collection is done, as is possible for volume scanning [9].

Although iOCT effectively increases the imaging speed by parallelizing data acquisition, ultimately, the sensitivity will be governed by the illuminating conditions when operating at the ANSI limit. This is a fundamental limitation of all SS-OCT systems with a beam size smaller than the allowable aperture. Multi-beam approaches can sidestep this limit when the beams are separated by more than the limiting aperture of the ANSI standards (>1 mm). In order to take advantage of this same idea with iOCT, one could consider dilating the line beam to exceed this aperture, although this is admittedly challenging.
Another limiting factor that governs the sensitivity in all Fourier-domain OCT systems is the number of data points in the A-scan. In this work, the number of points per A-scan was limited by the digitizer and the PCI bus to 4096 total. Hence, we were forced to trade off this number (N = 4096) with M (number of multiplexed points) to demonstrate a high degree of multiplexing. It is well known that the sensitivity of Fourier-domain techniques like SS-OCT is fundamentally related to the number of points in the A-scan [22]. As digitizer and PCI bus transfer speeds increase with new computing technology, the sensitivity of iOCT will correspondingly increase as well because we will be able to capture more data points to the extent that we are not limited by the coherence length of the source.

6. Conclusion

This manuscript describes highly parallel OCT imaging using iOCT with air-spaced VIPAs that we fabricated in a standard clean room at a very low cost. The number of multiplexed points could be tuned as desired to balance the effective imaging speed, ranging depth, and sensitivity. In this work we demonstrate an effective 3.2-MHz A-scan rate using a source with a sweep rate of only 200 kHz. This was achieved by multiplexing 16 interferograms onto a single sweep. Moreover, the presented system resolves many limitations in the previous work (e.g., larger ranging depth, higher sensitivity, no ghost images).

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