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Optical-domain subsampling for data efficient depth ranging in Fourier-domain optical coherence tomography

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Abstract: Recent advances in optical coherence tomography (OCT) have led to higher-speed sources that support imaging over longer depth ranges. Limitations in the bandwidth of state-of-the-art acquisition electronics, however, prevent adoption of these advances into the clinical applications. Here, we introduce optical-domain subsampling as a method for imaging at high-speeds and over extended depth ranges but with a lower acquisition bandwidth than that required using conventional approaches. Optically subsampled laser sources utilize a discrete set of wavelengths to alias fringe signals along an extended depth range into a bandwidth limited frequency window. By detecting the complex fringe signals and under the assumption of a depth-constrained signal, optical-domain subsampling enables recovery of the depth-resolved scattering signal without overlapping artifacts from this bandwidth-limited window. We highlight key principles behind optical-domain subsampled imaging, and demonstrate this principle experimentally using a polygon-filter based swept-source laser that includes an intra-cavity Fabry-Perot (FP) etalon.

OCIS codes: (170.4500) Optical coherence tomography; (140.3460) Lasers.

References and links


1. Introduction

Optical coherence tomography (OCT) images tissue microarchitecture at a high-resolution and without labeling, making it an attractive diagnostic tool in medical imaging. Early implementations of OCT operated at only several kHz imaging speeds, which limited its clinical utility. In 2004, higher-speed swept-wavelength OCT imaging was demonstrated at 10 kHz A-line rates [1]. Since that time, multiple new swept-wavelength technologies have been developed and laser speeds have increased to hundreds of kHz [2–6]. In parallel, the bandwidth of the electronic acquisition systems used to capture OCT signals has increased through adoption of higher-speed digitizers and higher bandwidth bus interfaces. With the successive improvements in these laser and acquisition subsystems, faster OCT imaging systems are being deployed into the clinic. The recent demonstration of MHz-range laser source designs [7], GHz-speed high bit-depth digitizers, and PCI-Express 3.0 bus interfaces will continue this trend over the next several years.

However, the bandwidth of state-of-the-art acquisition has approximately kept pace with increases in laser speed only if the imaging depth is held constant at the millimeter scale. In OCT, the required acquisition bandwidth scales with the product of the laser speed and imaging depth range. As new laser sources are demonstrated with multi-cm scale coherence lengths [6], new clinical and industrial applications of OCT based on simultaneous high-speed and multi-cm depth ranges can be envisioned. When the requirements of high-speed are combined with those of extended depth range imaging, current acquisition electronics are unable to accommodate the resulting signal bandwidth. This gap is sufficiently large that evolutionary improvements in acquisition electronics are likely not the solution. Instead, high-speed and extended depth range imaging may become feasible only with a more fundamental change in approach.

In this work, we propose, analyze, and demonstrate a method to dramatically reduce the acquisition bandwidth required for extended depth range imaging, and thereby enable high-speed and extended depth range OCT with current acquisition electronics. Our approach is based on modifying the optical sampling approach in OCT to achieve optical-domain subsampling of OCT interference fringes. Optical subsampling has been demonstrated
previously to extend the coherence length of source or reduce fringe decay [8–10], but has not
to our knowledge been demonstrated as a method for data-compressive ranging. We introduce
the concept of subsampling applied to Fourier-domain OCT fringe signals, describe the
optical-domain approach to subsampling, highlight key concepts and performance attributes
of optical subsampling, and finally present a preliminary demonstration of the optical
subsampling approach to acquisition bandwidth reduction.

2. Extended depth-range OCT

Because they operate with limited (mm-scale) depth ranges, existing clinical deployments of
OCT require tight control over the distance between the imaging catheter or microscope and
the tissue. In intravascular OCT, this distance is constrained by the intraluminal vessel
diameter such that a 5-6 mm imaging range is sufficient to maintain the tissue within the
imaging window [11]. For esophageal imaging, balloon catheters are used to center the
imaging optics so that the tissue falls within the imaging window on all sides of the probe
[12]. In skin and retinal imaging, the planar geometry of the tissue allows the tissue location
to be tightly controlled. However, many organs and sites feature irregular geometries that
cannot be as easily constrained. Extending the depth-range can make these sites accessible to
OCT.

For clarity, it is important to draw a distinction between the imaging depth range and the
imaging penetration into tissue. The former is defined by the OCT instrumentation and
defines the depth range over which signals are acquired, while the latter describes the furthest
depth in scattering tissue at which signals can be detected. Penetration depth is limited by
tissue optics to 1-2 mm regardless of the imaging depth range. Thus, in an extended depth-
range OCT embodiment, a typical A-line will contain regions of negligible signal both
superficial to the tissue surface, and at delays associated with locations deeper than 1-2 mm
beyond the tissue surface. Between these signal-absent regions will be the tissue signal region
(Fig. 1). Acquiring this full A-line is data inefficient because a large fraction of the
acquisition bandwidth is dedicated to the signal-absent (a scattering or attenuated) regions.
However, because the location of the tissue signal is not known a priori, a limited and
targeted acquisition of the depth range containing the tissue is practically challenging. The
acquisition bandwidth required for extended range imaging can be reduced by finding a way
to eliminate this inefficiency while preserving the ability to image over extended depth
ranges.

3. Subsampling of bandwidth limited signals

The sparsity of the extended depth range A-line illustrated in Fig. 1 suggests the use of data-
efficient sampling approaches. To explore this, we first review the relationship between the
A-line signal (Fig. 1) and the fringe signal that is directly measured by the OCT instrument
and from which the A-line signal is derived. In Fourier-domain OCT, coherent echo-delay
measurements are performed by measuring the interference fringe signal intensity between a
reference beam and light backscattered from the sample [1]. This interference fringe is
recorded as a function of the probing wavelength either through a swept-wavelength laser source and a photodetector, or a broadband source combined with a spectrometer. The interference fringe as a function of optical wavenumber, \( k = 2\pi/\lambda \), is related to the optical scattering across depth through a Fourier transformation. Because of this Fourier relationship, it can be appreciated that when the optical scattering signal is constrained to occupy a finite depth range, the resulting interference fringe signal will be the time-domain representation of a bandwidth limited (also termed bandpass) signal.

Approaches for sampling bandwidth limited signals have been studied extensively in communications and information theory [13–15]. Consider a bandwidth limited signal located at \( f_C \) with a bandwidth \( B \) (Fig. 2a,b). Nyquist sampling at \( 2f_C \) captures this signal fully, but is data inefficient because a large fraction of the detected bandwidth does not contain signal (Fig. 2b). Alternatively, because the signal is bandwidth limited, sampling the signal directly at twice its bandwidth, i.e., \( 2B \), can capture its information content (Fig. 2c). This approach is termed subsampling because it samples the signal at rates below twice the highest frequency content of the signal, \( 2f_C \). Higher frequencies appear in the baseband window through aliasing (Fig. 2d,e).

4. Signal overlap in subsampled Fourier-domain OCT fringes

To be effective in the context of imaging, subsampling must ensure that signals from each depth, within the penetration depth of tissue, can be measured independently from those at other depths, i.e., overlap artifacts must not compromise the resulting image. This requirement can be met straightforwardly when subsampling is applied to complex fringes, but is more challenging when applied to real fringes. To illustrate this, we generated a numerical OCT phantom structure (a simple smiley face), derived associated fringe signals from this phantom, and explored the effect of subsampling on the imaging (Fig. 3). First, we place the phantom at varying locations in depth (Fig. 3A). Next, we derive the associated fringes assuming continuous wavelength sampling across this depth range. We then subsampled the real-valued fringe data by a ratio of 1:4, and presented the compressed image (Fig. 3B, showing only the positive frequencies). Note that the non-circular mapping of signal
frequency to aliased frequency results in image overlap for most locations of the image. We then repeated this analysis but subsampled the complex fringe signal at a ratio of 1:8 (to give the same baseband window depth, the ratio was decreased by a factor of two because the signals are complex valued). In Fig. 3C, the complex subsampled images are presented. Note that the image is wrapped circularly, and signals never overlap onto itself for any depth location.

To achieve artifact-free imaging for arbitrary sample locations, it is necessary therefore to operate on complex OCT fringe data. Many approaches exist for extracting the complex optical interference signals in OCT. These have been used previously to separate signals from positive and negative delays and double imaging depth supported by a limited coherence length source. Options include dynamic methods based on modulating the signal using for example an acousto-optic frequency shifter [16], or physical methods based on generation of phase-shifted interference signals directly [17, 18].

Fig. 3. Real-valued and complex-valued signals are mapped differently into the aliased frequency space. For real-valued signals, signals located at varying locations (A) can induce distortions due to non-circular wrapping in the aliased image (B). For complex signals, wrapping is circular and overlap is avoided as long as the baseband window is large enough to contain the depth extent of the signal (C).

5. Electrical-domain subsampling in Fourier-domain OCT

The most straightforward implementation of subsampling is in electrical-domain, i.e., to maintain full RF bandwidth on all receivers but operate the digitization clock at a reduced rate. A conventional OCT receiver is shown in Fig. 4A and the electrical-domain subsampling receiver is illustrated in Fig. 4B. Here, the full interference fringe bandwidth is detected and transmitted to the digitizer, but is sampled at a rate of 2B rather than the Nyquist rate (2Fa) by the digitizer. By operating at a lower digitization rate, subsampling in the electrical-domain reduces the digital acquisition bandwidth required to capture the signal. However, by requiring full analog bandwidth, it also increases the noise proportionally by integrating noise across this large bandwidth into the aliased baseband window [19]. For some applications requiring a modest decrease in acquisition bandwidth, the associated noise increase might be an acceptable penalty to achieve a corresponding reduction in digital acquisition bandwidth. For more aggressive applications of subsampling, this noise penalty would be prohibitive. In this work, we describe a strategy that implements subsampling in the optical-domain such that acquisition bandwidth reductions can be achieved without proportional penalties in noise.
Fig. 4. Electrical-domain subsampled receiver designs. An electrical-domain subsampling receiver (B) must retain the full RF bandwidth of the conventional fully sampled received (A), but it utilizes a lower digitization clock rate ($2B$ vs. $2F_a$). The resulting digital acquisition bandwidth is reduced by the factor of ($F_a/B$) and the noise is increased by the same factor (assuming white noise).

6. Optical-domain subsampling in OCT

In addition to the electrical-domain, subsampling can be implemented in the optical-domain by limiting the wavelengths used to probe the sample, i.e., by probing the tissue with a set of discrete wavelengths rather than a continuously wavelength-swept source. Fringe signals of a continuously swept-wavelength and wavelength-stepped (subsampled) source are illustrated in Fig. 5. By stepping between wavelengths, multiple depth locations are aliased optically to the same fringe frequency, achieving an optical baseband compression of a large depth range. This fringe can be captured using reduced analog and reduced digital bandwidth receivers, and the noise bandwidth is less than that required for electrical-domain subsampling as a result (Fig. 6). Optical-domain subsampling therefore achieves the compression of subsampling without a proportional noise increase. Here, we describe some of the central properties of optical-domain subsampling. In the following section, preliminary demonstrations of optical-domain subsampled imaging are presented.

Fig. 5. An illustration of wavelength evolution and fringe signals generated from a continuously wavelength-swept laser (left) and an optical-domain subsampled laser source (right).

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6.1 Relationship between subsampled laser parameters and imaging parameters

An optical-domain subsampled source is described by its bandwidth and instantaneous linewidth as is the case for a continuously swept-wavelength source, but also requires definition of its discrete wavelength/wavenumber spacing, or its free spectral range (FSR). As in conventional swept-wavelength sources, the source bandwidth describes the axial resolution and the instantaneous line width describes the coherent-limited ranging depth [1]. The FSR in a subsampled source describes the size of the baseband aliased window, i.e., the depth extent of the baseband image. If we assume a constant spacing (in optical frequency) between wavelengths and complex demodulation, the FSR is related to the baseband depth, $D_S$, by Eq. (1):

$$FSR = \frac{c}{2nD_s} \quad (1)$$

Where $c$ is the speed of light and $n$ is the index of refraction of the sample medium. As described earlier, it is important to ensure that the baseband depth $D_S$ is greater than the depth extent of the signal ($D_I$ in Fig. 1) to eliminate signal overlap.

6.2 Signal loss due to higher order harmonics

The stepped optical fringe generated by optical-domain subsampling contains some power in higher-order harmonics. The information content of these harmonics is redundant to the baseband window, allowing them to be filtered prior to digitization. However, the power lost to these higher-order harmonics reduces the signal power in the baseband window and affects the measurement SNR. To explore this further, we analyzed the frequency content of a simulated step-wise interference fringe as a function of location (Fig. 7). These results demonstrate that there is a depth-dependent generation of higher-order harmonics. The depth-dependence can be explained by the varying magnitude of the step changes in associated interference fringes; greater magnitude of step changes occur at fringes that are closer to being critically sampled by the discrete set of wavelengths (Fig. 7e). This corresponds to signals located at $(m + 0.5)D_S$ where $D_S$ is the baseband window depth and $m$ is an integer; hence there is a maximum loss (3.8 dB) in baseband signal power here due to higher-order harmonics. Note that these results assume detection by a conventional low-pass filtered digitizer (see Fig. 6), and would be avoided if specialized detector circuits employing integrate and hold amplifier circuitry were to be used [20].
Fig. 7. Optical-domain subsampling induces a small periodic loss in baseband signal strength due to its stepwise nature and the resulting placement of signal power into higher orders. The signal variation is limited to 3.8 dB over the aliased baseband depth window. $m$ is an integer and $D_s$ is the baseband window depth.

6.3 Laser chirp in subsampled sources

Continuously swept-wavelength sources rarely sweep linearly in wavenumber-space, and thus require chirped acquisition clocks ($k$-clocking) [21] or interpolation of the fringes after digitization [22]. Generally, these approaches can be used to generate bandwidth limited depth point-spread functions for arbitrarily chirped sources. In optical-domain subsampling, two separate sources of laser chirp exist, one that can be handled by analogous routines and the other being fundamentally distorting. A subsampled laser source can be nonlinear-in-$k$ or nonlinear-in-time (or both) as illustrated in Fig. 8. Generally, optical-domain subsampling is compatible with nonlinear-in-time chirping through either non-uniform digitizer clocking or interpolation. However, interpolation and/or clocking cannot be used to address sources that are chirped in $k$, i.e., which feature a varying FSR. In this case, the underlying fringes do not repeat periodically with depth, and a depth-dependent distortion is directly induced. The applications of advanced approaches for spectral analysis of non-uniformly sampled signals may be applicable in such cases, but are beyond the scope of this work. Thus, for conventional optical-domain subsampling of bandwidth limited signals, it is important that the source have a constant FSR, but is not critical that each wavelength step occur at a fixed rate in time.
6.4 Optical subsampling in spectrometer-based Fourier-domain OCT

While optical-domain subsampling in this work has been described in a wavelength-stepped configuration with a time-varying source, it is also applicable to spectrometer-based systems using a continuous-wave comb source. The relationship between the source FSR and the imaging window is unchanged in this configuration. In a spectrometer-based system, nonlinear-in-time chirping is replaced by nonlinear-in-pixel chirping (but may be correctable by interpolation), while nonlinear-in-k chirping remains more disruptive. A more detailed analysis of the subsampled implementation in spectrometer-based OCT is beyond the scope of this work.

7. System Design and Construction

A preliminary optical-domain subsampled OCT system was constructed to allow key principles of depth-compressive imaging to be experimentally demonstrated.

7.1 Optical-domain subsampled laser

To generate the optical-domain wavelength stepped laser source, a continuously wavelength-swept laser based on a free space polygon mirror sliding filter was modified (Fig. 9). A free space fused silica Fabry-Perot (FP) etalon (Light Machinery) was inserted into the laser cavity to select discrete and linear-in-k wavelengths. The FSR of the FP was 80 GHz, providing a baseband window depth of 1.358 mm assuming a tissue index of \( n = 1.38 \). The FP finesse was specified as greater than 80 over a 100 nm spectral bandwidth centered at 1300 nm. The laser was implemented using a free-space optical circulator design, which allows smaller cavity lengths. The line width of the polygon-mirror based filter was designed to be approximately 0.21 nm, providing sufficient extinction of the neighboring FP modes while retaining a high laser duty cycle. A booster amplifier was placed outside the laser cavity to compensate for power losses induced by the insertion of the FP filter. Output power was measured at 46 mW. The laser was operated at 27 kHz and 5.4 kHz for experiments described in later sections.
7.2 Interferometer and data acquisition

The interferometer used in these experiments follows previously reported designs [1, 16]. An acousto-optic frequency shifter (AOFS) at 25 MHz was used in the reference arm to provide complex fringe demodulation. Trigger signals were generated from a fiber Bragg grating (FBG) with a center wavelength chosen with sufficient overlap to one of the FP transmission peaks. The FBG bandwidth of 42 GHz was sufficiently small relative to the FP’s FSR to ensure consistent trigger pulse generation from a specific FP transmission peak. Fringes were detected using balanced 80 MHz receivers (New Focus, 1817-FC). Electronic low-pass filtering at 50 MHz was implemented prior to digitization at 100 MS/sec. A single polarization channel was digitized and recorded. Fringe acquisition lengths were set at 1700 samples, providing approximately 60% acquisition duty cycle when running the source at 27 kHz.

8. Performance

8.1 Fringes

To examine the interference fringes, a separate Michelson interferometer was constructed. The A-line rate was reduced to 5.4 kHz, and the interferometer output was detected using a 200 MHz receiver (Thorlabs, BDB460C) without balanced detection. By operating at a lower A-line rate and using higher speed receivers, the higher frequency components and stepped nature of the fringe can be appreciated (Fig. 10). As the reference arm was translated, the fringe properties repeated periodically, confirming that subsampling was being performed in the optical domain.

Fig. 10. Interference fringe signals at two depths demonstrate optical-domain generation of baseband signals.
8.2 Laser coherence length

To measure the laser coherence length, we used the imaging interferometer described in Section 4.2. Fringe data were recorded from a fixed sample arm mirror while translating the reference arm over a 12.5 cm range (25 cm optical path variation). Fringe visibility calculated as the standard deviation of the fringe showed a single-pass coherence length of approximately 7.4 cm (Fig. 11). We note that without the intra-cavity FP filter, the laser coherence length was limited to several millimeters, demonstrating that inclusion of the fixed FP etalon can both force optical-domain subsampling and also contribute to significant extension of the laser coherence length.

8.3 Data compression and ranging

To examine the data compression provided by optical-domain subsampling, interference fringes of a sample mirror signal were acquired over a 15 mm optical path variation induced by translating the reference arm 7.5 mm. The interference fringes were recorded continuously as the reference mirror translated. Illustrated in Fig. 12 is the detection receiver along with the frequency content of the A-lines over a subset of this 7.5 mm translation. This Fourier transformed signal is divided into three regions; the baseband window region, the higher order harmonics, and the complex conjugate signals. The wrapping of the mirror signal in the baseband window and the circular nature of this wrapping can be appreciated; when the fringe signal reaches the edge of the baseband frequency window, it circularly wraps back to the beginning (opposite side) of the window at the next incremental depth. This pattern continues over the extent of the 7.5 mm translation range, and confirms baseband demodulation of signals over an extended depth range. This result is repeated in the higher order signal regions and in the complex conjugate domain.
Fig. 12. Experimental demonstration of optical-domain subsampled OCT. Interference fringes were acquired of a fixed sample while translating the reference arm mirror. The frequency content of the interference fringes demonstrates the wrapping of the mirror signal in the baseband window and presence of higher order powers.

8.4 Imaging

To validate that subsampling is applicable to OCT imaging, images of a finger and a rubber phantom were acquired with the same detection receiver outlined in Fig. 12. The A-line rate was 27 kHz. We used a modified microscope wherein a long focal length lens (f = 10 cm) was placed before the 2-axis galvometer mirrors instead of behind it (Fig. 13). The system advantageously allowed for an increased field-of-view (FOV) with dependence on how far away the sample was placed from the galvometer mirrors. Furthermore, the loose focusing of the optical beam increased the depth-of-focus so that imaging could be performed over an extended depth range at a reduced transverse resolution of ~96 μm.

Figure 14A shows the baseband window of one longitudinal cross-section of a finger. The higher order harmonic signals were cropped since they contained redundant information and were a consequence of using 100 MS/s digitizer in lieu of a slower digitizer. Since there was curvature in the finger, there is some discontinuity in the image due to aliasing of the tissue signal upon reaching the edge of the baseband window (arrows: location of aliasing of the surface of the sample). Interestingly, however, tiling identical copies of this baseband window lets us appreciate the continuity of the sample (Fig. 14B). In Fig. 14C, a depth cross section
reveals how these tiled baseband windows compile to form an *en face* image of the finger (arrow: junction between skin and finger nail). The wrapping caused by subsampling results in numerous depth slices being visualized in one *en face* cross-section. Theoretically, a surface finding algorithm can be employed to eliminate this effect, however, this redundant depth signals does not interfere with the interpretation of the image, and can help visualize the contour of the sample. Note that because the source FSR gives 1.358 mm of image depth, the signal has fully dropped below the system noise floor before signal from the surface reappears.

![Fig. 14. Cross-sectional images of a finger resting on a small breadboard, imaged with the subsampled OCT set-up. (A) Baseband window cross-section of the skin. Curvature of the sample causes wrapping of the surface at the location of the arrows. Scale bar: 500 μm. (B) Tiling the baseband window (outlined in yellow) allows for continuous visualization of the sample. Arrow: fixed frequency noise resulting from laser (C) *En face* view of the cropped/tiled image. Bar: location of longitudinal cross section in (A). Arrow: junction between the finger and the nail.]

The ability of subsampled OCT to support imaging over extended depth ranges is better demonstrated with a rubber phantom that is placed at a tilt so that the depth of the sample spans a range of 2 cm (Fig. 15A). An *en face* image of this set-up shows the numerous aliased
surface reflections of the rubber phantom, the resting metal post, and the small optical breadboard (Fig. 15B). A movie showing the full en face data set is provided to communicate the nature of these subsampled images.

Fig. 15. (A) Rubber phantom resting against a metal post on a small optical breadboard. The tilted rubber phantom spans 2 cm in depth [Media]. (B) An en face cross section of the rubber, post, and breadboard. Aliases of the tilted rubber phantom from different depth planes into this one make it possible to visualize numerous surface reflections.

9. Conclusion

In this paper, we have described optical-domain subsampling as a means for imaging at high-speeds and over extended depth ranges while minimizing the required analog and digital acquisition bandwidth. The approach is based on optical aliasing of interference signals to a baseband frequency window, and in principle does not increase the noise floor. Critical concepts including noise, signal loss to higher-harmonics, receiver design, and laser chirping were analyzed within the context of optical-domain subsampling. To validate the principle of optical-domain subsampling, we constructed a prototype optical-domain subsampled laser by inserting a fixed FP etalon into a continuously wavelength-swept laser cavity and demonstrated baseband compression of a mirror signal over a 7.5 mm translation. We imaged a finger and a rubber phantom to demonstrate how subsampling can be incorporated in OCT imaging. A detailed evaluation of the noise properties of the subsampled approach and elimination of noise through building new iterations of subsampled lasers will be the focus of future work. Optical-domain subsampling approaches have the potential to enable the application of OCT imaging techniques to a new set of applications in clinical medicine and industrial settings.

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