Extended bandwidth wavelength swept laser source for high resolution optical frequency domain imaging

SAHAR HOSSEINZADEH KASSANI,1,2,* MARTIN VILLIGER,1,2 NÉSTOR URIBE-PATARROYO,1,2 CHANGSU JUN,1,2,3 REZA KHAZAEINEZHAD,1,2 NORMAN LIPPOK,1,2 AND BRETT E. BOUMA1,2,4

1Harvard Medical School, Boston, Massachusetts 02115, USA
2Wellman Center for Photomedicine, Massachusetts General Hospital, Boston, Massachusetts 02114, USA
3Currently with Advanced Photonics Research Institute, GIST, Gwangju 61005, South Korea
4Institute for Medical Engineering and Science, Massachusetts Institute of Technology, Cambridge, Massachusetts 02142, USA
*skassani@mgh.harvard.edu

Abstract: Improving the axial resolution by providing wider bandwidth wavelength swept lasers remains an important issue for optical frequency domain imaging (OFDI). Here, we demonstrate a wide tuning range, all-fiber wavelength swept laser at a center wavelength of 1250 nm by combining two ring cavities that share a single Fabry-Perot tunable filter. The two cavities contain semiconductor optical amplifiers with central wavelengths of 1190 nm and 1292 nm, respectively. To avoid disturbing interference effects in the overlapping spectral region, we modulated the amplifiers in order to obtain consecutive wavelength sweeps in the two spectral regions. The two sweeps were fused together in post-processing to achieve a total scanning range of 223 nm, corresponding to 3.3 µm axial resolution in air. We confirm improved image quality and reduced speckle size in tomograms of swine esophagus ex vivo, and human skin and nailbed in vivo.

© 2017 Optical Society of America

OCIS codes: (110.0110) Imaging systems; (140.3600) Lasers, tunable; (110.4500) Optical coherence tomography; (070.0070) Fourier optics and signal processing.

References and links
1. D. Huang, E. A. Swanson, C. P. Lin, J. S. Schuman, W. G. Stinson, W. Chang, M. R. Hee, T. Flotte, K. Gregory, C. A. Puliafito, and et, “Optical coherence tomography,” Science 254(5035), 1178–1181 (1991).
2. U. Morgner, W. Drexler, F. X. Kärntner, X. D. Li, C. Pirris, E. P. Ippen, and J. G. Fujimoto, “Spectroscopic optical coherence tomography,” Opt. Lett. 25(2), 111–113 (2000).
3. S. R. Chinn, E. A. Swanson, and J. G. Fujimoto, “Optical coherence tomography using a frequency-tunable optical source,” Opt. Lett. 22(5), 340–342 (1997).
4. R. Leitgeb, C. Hitzenberger, and A. Fercher, “Performance of fourier domain vs. time domain optical coherence tomography,” Opt. Express 11(8), 889–894 (2003).
5. M. Choma, M. Sarunic, C. Yang, and J. Izatt, “Sensitivity advantage of swept source and Fourier domain optical coherence tomography,” Opt. Express 11(18), 2183–2189 (2003).
6. J. F. de Boer, B. Cense, B. H. Park, M. C. Pierce, G. J. Tearney, and B. E. Bouma, “Improved signal-to-noise ratio in spectral-domain compared with time-domain optical coherence tomography,” Opt. Lett. 28(21), 2067–2069 (2003).
7. W. Drexler and J. G. Fujimoto, “State-of-the-art retinal optical coherence tomography,” Prog. Retin. Eye Res. 27(1), 45–88 (2008).
8. T. Xie, G. Liu, K. Kreuter, S. Mahon, H. Colt, D. Mukai, G. M. Peavy, Z. Chen, and M. Brenner, “In vivo three-dimensional imaging of normal tissue and tumors in the rabbit pleural cavity using endoscopic swept source optical coherence tomography with thoracoscopic guidance,” J. Biomed. Opt. 14(6), 064045 (2009).
9. M. J. Suter, B. J. Vakoc, P. S. Yachimski, M. Shishkov, G. Y. Lauwers, M. Mino-Kenudson, B. E. Bouma, N. S. Nishioaka, and G. J. Tearney, “Comprehensive microscopy of the esophagus in human patients with optical frequency domain imaging,” Gastrointest. Endosc. 68(4), 745–753 (2008).
10. G. Guagliumi and V. Sirbu, “Optical coherence tomography: high resolution intravascular imaging to evaluate vascular healing after coronary stenting,” Catheter. Cardiovasc. Interv. 72(2), 237–247 (2008).
11. S. H. Yun, G. Tearney, J. de Boer, and B. Bouma, “Motion artifacts in optical coherence tomography with frequency-domain ranging,” Opt. Express 12(13), 2977–2998 (2004).
12. S. Yun, G. Tearney, J. de Boer, N. Itifimia, and B. Bouma, “High-speed optical frequency-domain imaging,” Opt. Express 11(22), 2953–2963 (2003).
13. B. Potsaid, V. Jayaraman, J. G. Fujimoto, J. Jiang, P. J. S. Heim, and A. E. Cable, “MEMS tunable VCSEL light source for ultrahigh speed 60kHz - 1MHz axial scan rate and long range centimeter class OCT imaging,” in J. A. Izatt, J. G. Fujimoto, and V. V. Tuchin, eds. (International Society for Optics and Photonics, 2012), p. 82130M.
14. M. Kuznetsov, A. Tiaia, B. Johnson, and D. Flanders, “Compact ultrafast reflective Fabry-Perot tunable lasers for OCT imaging applications,” in J. A. Izatt, J. G. Fujimoto, and V. V. Tuchin, eds. (International Society for Optics and Photonics, 2010), p. 75541F.
15. W. Y. Oh, S. H. Yun, G. J. Tearney, and B. E. Bouma. “115 kHz tuning repetition rate ultrahigh-speed wavelength-swept semiconductor laser,” Opt. Lett. 30(23), 3159–3161 (2005).
16. R. Huber, M. Wojtkowski, and J. G. Fujimoto, “Fourier Domain Mode Locking (FDML): A new laser operating regime and applications for optical coherence tomography,” Opt. Express 14(8), 3225–3237 (2006).
17. I. Grulkowski, J. J. Liu, B. Potsaid, V. Jayaraman, C. D. Lu, J. Jiang, J. S. Duke, and J. G. Fujimoto, “Retinal, anterior segment and full eye imaging using ultrahigh speed swept source OCT with vertical-cavity surface emitting lasers,” Opt. Express 3(11), 2733–2751 (2012).
18. R. Huber, M. Wojtkowski, K. Taira, J. Fujimoto, and K. Hsu, “Amplified, frequency swept lasers for frequency domain reflectometry and OCT imaging: design and scaling principles,” Opt. Express 13(9), 3513–3528 (2005).
19. Y. Shimada, A. Sadr, M. F. Burrow, J. Tagami, N. Ozawa, and Y. Sumi, “Validation of swept-source optical coherence tomography (SS-OCT) for the diagnosis of occlusal caries,” J. Dent. 38(8), 655–665 (2010).
20. F. Lexer, C. K. Hitzenberger, A. F. Fercher, and M. Kulhavy, “Wavelength-tuning interferometry of intracoureal distances,” Appl. Opt. 36(25), 6548–6553 (1997).
21. Y. Wang, Y. Zhao, J. S. Nelson, Z. Chen, and R. S. Windeler, “Ultrahigh-resolution optical coherence tomography by broadband continuum generation from a photonic crystal fiber,” Opt. Lett. 28(3), 182–184 (2003).
22. R. Leitgeb, W. Drexler, A. Unterhuber, B. Hermann, T. Bajraszewski, T. Le, A. Stingl, and A. Fercher, “Ultrahigh resolution Fourier domain optical coherence tomography,” Opt. Express 12(10), 2156–2165 (2004).
23. M. Wojtkowski, V. Srinivasan, T. Ko, J. Fujimoto, A. Kowalczyk, and J. Duker, “Ultrahigh-resolution, high-speed, Fourier domain optical coherence tomography and methods for dispersion compensation,” Opt. Express 12(11), 2404–2422 (2004).
24. L. Liu, J. A. Gardecki, S. K. Nadkarni, J. D. Toussaint, Y. Yagi, B. E. Bouma, and G. J. Tearney, “Imaging the subcellular structure of human coronary atherosclerosis using micro-optical coherence tomography,” Nat. Med. 17(8), 1010–1014 (2011).
25. W.-Y. Oh, S.-H. Yun, G. J. Tearney, and B. E. Bouma, “Wide Tuning Range Wavelength-Swept Laser With Two Semiconductor Optical Amplifiers,” IEEE Photonics Technol. Lett. 17(3), 678–680 (2005).
26. H. Kim, J. Lim, J. Ha, and W. Jang, “Broadband wavelength swept source combining a quantum dot and a quantum well SOA in the wavelength range of 1153–1360 nm,” Electron. Lett. 49(19), 1205–1206 (2013).
27. M. Y. Jeon, Y. Wang, Z. Chen, and A. F. Fercher, “High-speed and wide bandwidth Fourier domain mode-locked wavelength swept laser with multiple SOAs,” Opt. Express 16(4), 2547–2554 (2008).
28. D. C. Adler, Y. Chen, R. Huber, J. Schmitt, J. Connolly, and J. G. Fujimoto, “Three-dimensional endomicroscopy using optical coherence tomography,” Nat. Photonics 1(12), 709–716 (2007).
29. C. Jun, M. Villiger, W.-Y. Oh, and B. E. Bouma, “All-fiber wavelength sweep ring laser based on Fabry-Perot filter for optical frequency domain imaging,” Opt. Express 22(21), 25805–25814 (2014).
30. Y. Nakazaki and S. Yamashita, “Fast and wide tuning range wavelength-swept fiber laser based on dispersion tuning and its application to dynamic FBS sensing,” Opt. Express 17(10), 8310–8319 (2009).
31. L. A. Kranendonk, X. An, A. W. Caswell, R. E. Herold, S. T. Sanders, R. Huber, J. G. Fujimoto, Y. Okura, and Y. Urata, “High speed engine gas thermometry by Fourier-domain mode-locked laser absorption spectroscopy,” Opt. Express 15(23), 15115–15128 (2007).
32. S. Yun, G. Tearney, J. de Boer, and B. Bouma, “Removing the depth-degeneracy in optical frequency domain imaging with frequency shifting,” Opt. Express 12(20), 4822–4828 (2004).
33. W. Wieser, B. R. Biedermann, T. Klein, C. M. Eigenwillig, and R. Huber, “Multi-megahertz OCT: High quality 3D imaging at 20 million A-scans and 4.5 GVoexels per second,” Opt. Express 18(14), 14685–14704 (2010).
34. A. Bilenca, S. H. Yun, G. J. Tearney, and B. E. Bouma, “Numerical study of wavelength-swept semiconductor ring lasers: the role of refractive-index nonlinearities in semiconductor optical amplifiers and implications for biomedical imaging applications,” Opt. Lett. 31(6), 760–762 (2006).
35. S. Yun, G. Tearney, J. de Boer, N. Itifimia, and B. Bouma, “High-speed optical frequency-domain imaging,” Opt. Express 11(22), 2953–2963 (2003).
36. Y. Piedderrière, J. Cariou, Y. Guerin, B. Le Jeune, G. Le Brun, and J. Lortr, “Scattering through fluids: speckle size measurement and Monte Carlo simulations close to and into the multiple scattering,” Opt. Express 12(1), 176–188 (2004).
1. Introduction

Optical coherence tomography (OCT) is a non-invasive, high-resolution, cross-sectional imaging modality based on the detection of light back-scattered from a sample [1]. The development of Fourier-domain OCT, including spectral-domain OCT (SD-OCT) [2] and frequency-domain OCT (OFDI or SS-OCT) [3], has led to substantial improvements in detection sensitivity [4–6]. This imaging technique has now found widespread application in ophthalmology while expanding into cardiology, oncology and gastroenterology [7–10].

While SD-OCT has been widely adopted for ophthalmic applications, where the spectral range of mature silicon-based cameras is well suited, frequency-domain systems have proven less susceptible to fringe washout in applications where motion is significant [11, 12]. Multiple distinct scanning filter designs have been implemented that offer high imaging speed, integration with optical fiber devices, and good image penetration depth by operating in the 1300 nm region [13–16].

Recent developments in wavelength swept laser design for OFDI have increased the instantaneous coherence length [17], however, the axial resolution has stagnated around ~10 µm [18, 19]. In order to achieve higher axial resolution with coherence gating, wider sweep range light sources are required since the axial resolution is inversely proportional to the sweep range:

$$\delta z = \frac{2 n \lambda^2}{\pi n \Delta \lambda},$$

where $\lambda$ is the center wavelength, $\Delta \lambda$ is the full-width at half-maximum (FWHM) of the integrated spectral envelope, a Gaussian-shaped spectrum is assumed, and $n$ is the group refractive index of the sample [20]. Although a number of broadband light sources have been developed to improve axial resolution in time domain and SD-OCT [21–24], only a few broadband high-speed wavelength-swept light sources have been reported [25–28].

Semiconductor gain media can be easily incorporated into compact laser cavity designs and have been developed for many different spectral bands. However, the tuning range of wavelength-swept lasers has been limited, especially at rapid scan rates, due to the limited gain bandwidth of individual semiconductor gain materials. Although quantum-dot-based SOAs are emerging to provide very wide gain bandwidths [26], further development is required before they find widespread application. An alternative approach to achieving an increased tuning range relies on the combination of two or more gain media whose gain spectra are distinct. Several strategies have been proposed to this end. A system using two semiconductor optical amplifiers (SOAs) in ring cavities sharing one polygon scanning filter has been demonstrated at a tuning range of 145 nm [25]. The development of a Fourier domain mode-locked wavelength-swept laser with the use of a Fabry-Perot (FP) filter and the parallel configuration of two SOAs has also been reported, with a tuning range of 160 nm resulting in an axial resolution of 6.6 µm in air [27]. However, these efforts have encountered significant difficulties, including: large cavity loss, and interference of overlapping spectra.

In our previous work [29], we demonstrated an all-fiber wavelength-swept laser based on an FP filter and a short-length cavity design for OFDI. The most significant characteristics of this method were simplicity of assembly, low cost of materials, and robust operation without polarization control. The proposed inexpensive design was capable of being reconfigured and customized for different performance metrics and simple enough to be adopted broadly in order to facilitate OCT research. A sweeping speed of 50-300 kHz with 110-150 nm sweep range and 40-100 mW of average output power at 1.3 µm was demonstrated.

Here, we extended this design by combining two ring cavities having separate SOAs but sharing a single FP filter to achieve high resolution OFDI. The SOAs had central wavelengths of 1190 nm and 1292 nm and the FP filter had a resonance frequency of 50 kHz. The achieved tuning range of 223 nm and axial resolution of 3.3 µm in air corresponds, to the best
of our knowledge, to the widest bandwidth among wavelength swept lasers reported to date. The primary insight that enabled this result was to avoid interference of overlapping spectra by using consecutive wavelength sweeps in the two spectral regions by modulating the two amplifiers in synchrony with the FP driving signal. Interferograms corresponding to the two sweeps were fused in post-processing. We confirmed the superior image quality of tomograms acquired with this extended bandwidth source. In addition to OCT, the extended bandwidth wavelength-swept laser may find application in sensing [30] or spectroscopy [31].

2. Methods

2.1 Extended bandwidth source configuration

Figure 1(a) shows the experimental configuration for the extended bandwidth wavelength swept laser. The 3-dB bandwidth of amplified spontaneous emission (ASE) of the first SOA (SOA-1: Innolume Inc.) was 90 nm with a center wavelength of 1190 nm. The second SOA, employed in separate ring cavity, (SOA-2: Thorlabs Inc.) had a 3-dB bandwidth of amplified spontaneous emission (ASE) of 92 nm and a 1292 nm center wavelength. Figure 1(b) shows the optical spectra of amplified spontaneous emission of the two SOAs individually and combined. In our previous work [29], we realized the importance of unidirectional isolation in the ring cavity when it includes any reflective component like the FP filter. In order to meet this objective, in addition to the isolators contained in the SOA packages, further isolation was provided by adding two isolators in the cavities to prevent amplification of backward circulating light. The arrows in the Fig. 1(a) indicate the direction of isolation in the ring cavities.

The FP filter (Micron Optics, FFP-TF) inside the ring cavities had a free spectral range (FSR) of ~210 nm at 1300 nm, a linewidth of ~0.2 nm, and a resonance frequency of 50 kHz. The two SOAs were modulated with a ~17% duty cycle square wave and were synchronized with the sinusoidal wave driving the FP in such a way that each SOA was turned on only during the positive wavelength sweep one after the other, as illustrated in Fig. 1(c). We, thus, were overdriving the FP filter to obtain a total sweep range exceeding the FSR. However, the SOA’s consecutive turning on and off prevented simultaneous lasing through multiple transmission orders of the filter [29]. The outputs of the two SOAs and the FP filter were combined with 50:50 couplers to define two coupled ring-cavities. The laser ring cavity output was split in two parts and amplified with a second set of SOAs, having the same
characteristics and modulations as the SOAs employed in the master oscillator. Two polarization controllers were manually adjusted to align the polarization state of the cavity output with the gain axes of the two SOAs in the amplification stage. The amplified laser light then was combined using a customized two-channel wavelength division multiplexer (WDM) with design wavelengths of 1190 nm for channel 1 and 1305 nm for channel 2, (Thorlabs Inc.). The average output power of the laser was measured as ~30 mW, the duty cycle was 34%, and the effective repetition rate was 25 kHz for each spectral band.

2.2 Source bandwidth and axial resolution estimation

Figure 2(a) displays the optical spectrum of each ring separately as well as the combined output of the two cavities. By carefully adjusting the duty cycle and precise timing of each SOA modulation signal with respect to the sinusoidal driving signal of the FP filter, a total scanning range of 223 nm (1140 nm to 1363 nm), with no dip at the joining point, was obtained. The small fluctuation for short wavelengths in band-1 was due to the initial fluctuation in ASE of SOA-1. The bandwidth-limited axial resolution calculated by inverse Fourier transformation of the source spectrum was 3.2 µm as shown in Fig. 2(b), which was computed with no spectral filtering.

2.3 Imaging system and calibration

Figure 3(a) shows the schematic of an optical frequency domain imaging system. The output from the extended bandwidth laser source was fed into a fiber-based Mach-Zehnder interferometer. In each arm, a 50/50 coupler directed the light towards the sample and reference mirror, respectively. Wideband optical couplers (Thorlabs Inc., 1300+/−100 nm) were utilized in this set up. The reference arm signal was modulated with an acousto-optic modulator at 50 MHz to remove depth-degeneracy [32]. The sample and reference signal were combined with a polarization diverse mixer (Finisar Corp.) and detected with two balanced receivers (Exalos), suppressing the DC level and common mode noise. The electric signal of each detector, corresponding to the rapidly wavelength-swept interference signal, was digitized at 200 MHz (Alazar Tech.). The acquisition of 4960 sample points of each pair of sweeps was triggered by the reflection of a fiber-Bragg grating at 1320 nm. Phase-locked loops ensured accurate synchronicity between the laser driving signals, the acousto-optic modulator, and the acquisition clock. Figure 3(b) displays the time trace of the laser output and the interference signal. Note that by adding a delay line, the two spectra could be arranged adjacent to each other in the temporal domain. The final output repetition rate could then be doubled by use of a Mach-Zehnder interleaver [33]. It would also be possible to recover the repetition rate of the FP filter by operating one SOA with a positive wavelength.
sweep and the second with a negative wavelength sweep. However, nonlinear effects in the SOA transfer optical energy from shorter toward longer wavelengths, favoring a positive sweep direction [34] and resulting in higher output power and a narrower instantaneous linewidth.

The two spectral regions were swept consecutively but were recorded in a single time trace. Accordingly, two mapping functions $f_1: t_1 \rightarrow k$ and $f_2: t_2 \rightarrow k$ are needed to interpolate the two relevant periods of the trace and recover the fringe signal as a function of linear wavenumber $k$. Although the absolute wavenumbers are not critical, the relative offset between the two spectral regions needs to be precisely known to enable their fusion into a single continuous interference signal. We acquired a data set with an attenuated mirror reflection in the sample arm, and varied the optical path length by moving the reference mirror during the acquisition. Treating each spectral region independently, we recovered the interpolation functions that linearize the fringe signals in the wavenumber domain and digitally compensated for dispersion mismatch in the interferometer [35]. Figures 4(a) and 4(b) show the resulting tomograms of the calibration data sets.

By default, the interpolation operation preserves the (distinct) number of sampling points present in each spectral region ($N_1 = 1424$, $N_2 = 1144$). Although both interpolated traces are linear in $k$, the spacing of their sampling $\Delta k_{1,2}$ and spectral range $N_{1,2} \cdot \Delta k_{1,2}$ are in general different, resulting in unequal axial scaling, off by a factor $\alpha$, as illustrated with dashed lines in Fig. 4(a) and (b). Furthermore, the slight difference in the group delay due to the dispersion imbalance of the interferometer resulted in an axial offset, $\beta$, between the two signals. Interpolating the more coarsely sampled spectral region on a finer grid, adding a phase ramp to adjust the axial offset, and zero-padding both spectra to $N_1 + N_2$ points resulted in perfectly matching peak positions for the tomograms of both regions as shown in Figs. 4(c) and 4(d). Next, observing the phase difference between the signal peaks of the two tomograms for the varying reference arm positions revealed the relative offset of the two spectra in $k$. At the correct offset, $n_\text{ao}$, this phase difference should be constant across all peak positions. Monitoring the phase variance across all peak positions yielded a clearly defined minimum, as shown in Fig. 4(e). Lastly, compensation with the constant scalar phase value $\phi_\text{ao}$, caused by the evolving carrier phase of the acousto-optic modulator and potentially additional phase offsets, enabled coherent addition of the complex valued tomograms, resulting in constructive interference of the peak signal, and destructive interference at the peak shoulders. Figure 4(f)
depicts the correctly interpolated, dispersion compensated, and adjusted fringe signals of an individual peak. Table 1 summarizes the required steps for the calibration and fusion process, starting from the two spectra of the calibration data set, \( S_1(m, z_p) \) and \( S_2(n, z_p) \), as functions of their wavenumber sample indices \( m, n \) and the calibration mirror position \( z_p \). These indices correspond to the sampled wavenumbers via \( k = k_{end} - m\Delta k_i = k_{end} - k_{os} - n\Delta k_2 \), where \( k_{end}, k_{os}, \) and \( \Delta k_{i,2} \) are originally unknown. The goal of the calibration procedure is to find the ratio between \( \Delta k_i \) and \( \Delta k_2 \), as well as the relative offset of the spectra \( k_{os} \) in terms of \( \Delta k_2 \), assuming that \( \Delta k_2 \geq \Delta k_i \).

![Fig. 4. Mapped and dispersion compensated depth-scans for band-1 (a), and band-2 (b), respectively. Same depth-scans, after additional calibration and zero-padding to \( N_1 + N_2 \) points resulting in equal depth positions for band-1 (c) and band-2 (d). (e) Standard deviation of the phase difference between band-1 and band-2 across all peak positions as a function of the relative offset in wavenumber indices between the two bands. (f) Synthesized fringe signals of an individual peak after applying correct wavenumber offset and an additional scalar phase value.](image-url)
Table 1. Calibration and fusion process

| **Input:** $S_1(m, z_p)$ and $S_2(n, z_p)$ | $m \in [0, N_1-1]$, $n \in [0, N_2-1]$ $N_{1,2}$: Number of samples $z_p$: Mirror position |
| **Output:** $\alpha, \beta, n_{os}, \phi_{os}$ | $\alpha = \Delta k_1/\Delta k_2$ with $\Delta k_2 \geq \Delta k_1$ $\beta$: Group delay offset $\phi_{os}$: Phase offset $k_{os} = n_{os} \Delta k_2$ |

1. Compute tomograms:

   $t_1(p, z_p) = \sum_{m=0}^{N_1-1} S_1(m, z_p) \exp\left(-i2\pi \frac{mp}{N_{FFT}}\right)$

   $t_2(q, z_p) = \sum_{n=0}^{N_2-1} S_2(n, z_p) \exp\left(-i2\pi \frac{nq}{N_{FFT}}\right)$

2. Find axial scaling $\alpha$ and offset $\beta$:

   $p_p = \alpha (q_p + \beta)$, $\alpha \leq 1$ $p_p, q_p$: Axial positions of the reflector peak signal

3. Compute adjusted $t_1'$ after interpolating $S_1$ on coarser $\Delta k_1'$ = $\Delta k_1/\alpha$ and adding a phase ramp to compensate the axial offset:

   $t_1'(p, z_p) = \sum_{m=0}^{N_1-1} S_1'(m, z_p) \exp\left(-i2\pi \frac{mp}{N_{FFT}}\right)$ where $S_1'(k') = S_1(k') \exp\left(i2\pi \frac{\beta m}{N_{FFT}}\right)$, $k' = m \frac{\Delta k_1}{\alpha}$

4. Compute candidate $t_1'$ for all possible offsets $n_{os}$:

   $t_1'(q, z_p, n_{os}) = \sum_{n=0}^{N_2-1} S_2(n, z_p) \exp\left(-i2\pi \frac{(n + n_{os})q}{N_{FFT}}\right)$ $n_{os}$: Relative offset, in integer sampling points, between the two spectra

5. Calculate phase variance across all peak positions and find best

   $n_{os}, \hat{n}_{os} = \min_{n_{os}} \left[ \arg\left(t_1'(p, z_p, n_{os}) \right) \arg\left(t_1'(q, z_p, \hat{n}_{os}) \right) \right]$ $\hat{n}_{os}$: Optimum offset $t_1'(p, z_p, \hat{n}_{os})$, $t_1'(q, z_p, \hat{n}_{os})$: Tomograms at the reflector peak signal

6. Calculate constant scalar phase offset:

   $\phi_{os} = \left[ \arg\left(t_1'(p, z_p) \right) \arg\left(t_1'(q, z_p, \hat{n}_{os}) \right) \right]$ $< >$: Mean over all $z_p$

7. Compute fused spectrum:

   $S_{os}(v) = S_1'(v) + S_2(v - \hat{n}_{os}) e^{i\phi_{os}}$ $v \in [0, N_{FFT}]$, $v = m$ for $S_1'$, and $v = n$ for $S_2$, and $S_1'(w) = 0$ for $w \geq N_1$ and $S_2(w) = 0$ for $w < 0$ and $w \geq N_2$

3. Results and discussions

3.1 Axial point spread function and resolution measurement

Figure 5(a) displays the axial point spread functions (PSFs) for various axial peak positions for the two individual and the combined spectra (squared norm of the complex valued tomogram). Figure 5(b) shows normalized and zoomed-in versions of one of the peaks, illustrating the coherent addition and resulting improvement in the axial resolution. Figure 5(c) presents the full-width at half-maximum (FWHM) of the axial PSFs for the individual and the combined spectra as a function of axial peak position, evidencing an axial resolution of 3.3 µm in air. No spectral shaping or filtering of the source spectrum has been done in this work. The experimentally measured axial resolution is in good agreement with the transform-limited value of 3.2 µm estimated in section 2.2. Interestingly, the FWHM of the combined source showed best performance in the center of the depth range, whereas the PSFs of the single sources had a narrower FWHM away from the center. The FWHM does not consider
side-lobes, and the single sources indeed exhibited more pronounced side-lobes away from the center, which can have the effect of narrowing the FWHM of the central lobe. The combined source, on the other hand, exhibited quite uniform side-lobes along the entire imaging range.

We found that the $k$-space calibration was, not surprisingly, sensitive to drift or intentional mis-adjustment of the filter sweep waveform. When the emitted laser spectrum changed significantly enough to be readily apparent under visualization using an optical spectrum analyzer, it was necessary to adjust the filter offset voltage to restore spectral shape and performance of the calibration. Despite having no temperature control, the FP was stable enough for the calibration to remain valid over the course of six hours.

Fig. 5. (a) Axial point spread functions (PSFs) for various axial peak positions for the two individual and the combined spectra, (b) Zoomed-in version of one of the peaks (c) full-width at half-maximum (FWHM) of the axial PSFs for the individual and the combined spectra

### 3.2 High resolution OCT imaging using extended bandwidth source

To evaluate the extended bandwidth wavelength swept laser, OCT imaging was performed with the system presented in section 2.3. OCT images of swine esophagus ex vivo corresponding to Band-1, Band-2 and the extended bandwidth laser are presented in Figs. 6 (a), (b) and (c), respectively.
The layered structure of the esophagus, submucosal structures and blood vessels can be better appreciated with the combined spectra as shown in Fig. 6(c), than in the tomograms corresponding to the individual spectral bands. No averaging/filtering was performed in any of the tomograms and an objective lens with 13 µm spot size (1/e² of beam diameter in the field of focus) was used for all images. Hence, the axial resolution significantly exceeds the lateral resolution. The achieved improvement in axial resolution could likely be further appreciated when paired with higher lateral resolution. Ultrasound gel was applied on the tissue surface to suppress specular reflection, although one remaining reflection can be seen in the images. In order to evaluate the speckle size of each tomogram, the dashed square region in each tomogram was selected as seen in Fig. 6(d) and then the lateral mean of the axial autocorrelation of each region was computed. The speckle size of each tomogram was evaluated as the full-width at half-maximum (FWHM) of the autocorrelation peaks [36] as illustrated in Fig. 6(e). As expected, the tomogram obtained with the extended bandwidth source showed a two-fold speckle size reduction in comparison with each of the single spectral bands.

Figure 7 shows several examples of OCT tomograms, comparing acquisition with individual spectral band and the extended bandwidth source. All tomograms are single, non-averaged B-scans consisting of 1024 A-scans.
Fig. 7. (a), (b) OCT tomograms of human skin (back of a hand) in vivo for a single band (a) and extended bandwidth laser (b), respectively. (c), (d) OCT images of swine esophagus ex vivo for a single band (c) and extended bandwidth laser (d). (e), (f) Images of a human finger (nailbed) in vivo for a single band (e) and extended bandwidth laser (f). Insets: 2x magnified view of the dashed regions.

All three examples demonstrate the improved image quality of the extended bandwidth laser source. Using the extended bandwidth laser for imaging human skin in the back of a hand in vivo, allowed the visualization of small vessels, highlighted by the dashed region and shown with larger magnification in the insets in Fig. 7(a) and (b). These features are less apparent using a single band source. Figures 7(c) and 7(d) show tomograms of a human finger (nailbed) acquired in vivo. Morphological structures such as dermis, epidermis, sweat ducts, stratum corneum, epidermal-dermal junction, and the nail structure can be seen more clearly with the extended bandwidth laser arising from improved axial resolution and benefiting from higher signal-to-noise ratio by adding two sources. Figures 7(e) and 7(f) show tomograms of swine esophagus ex vivo. The layered structure of the esophagus is difficult to observe in the tomogram acquired with a single band laser, but readily appreciated with the extended bandwidth source.

4. Summary

In summary, we have demonstrated an all-fiber, extended bandwidth wavelength-swept laser based on a single Fabry-Perot filter and two short-length ring cavities for OFDI, which enables imaging of biological tissues with ultrahigh resolution. Our results demonstrate a total
sweep range of 223 nm corresponding to an axial resolution of 3.3 µm in air which is the highest axial resolution achieved among wavelength-swept frequency-domain OCT systems, to the best of our knowledge. The design can be further optimized with respect to the sweep rate utilizing a buffered delay line for increasing the duty cycle. By integrating the laser with an OFDI system, we demonstrated high quality imaging of human skin, nailbed and swine esophagus. Results indicated a two-fold reduction in speckle size of the images. Small size structures could be clearly identified with respect to a single band laser. One limitation of the current design is the increased cost of the bill of materials due to the second SOA in the master oscillator. Further, since our target applications have been for imaging through rotational catheters where insertion losses may be high, we have elected to include booster amplifiers, further increasing cost.

Funding

National Institute of Biomedical Imaging and Bioengineering of the National Institutes of Health, award P41 EB015903, and by Terumo Corporation.