Swept source optical coherence tomography using an all-fiber 1300-nm ring laser source

Michael A. Choma
Duke University
Department of Biomedical Engineering
136 Hudson Hall
Durham, North Carolina 27708
E-mail: mac32@duke.edu

Kevin Hsu
Micron Optics, Incorporated
Atlanta, Georgia 30345

Joseph A. Izatt
Duke University
Department of Biomedical Engineering
Durham, North Carolina 27708

Abstract. The increased sensitivity of spectral domain optical coherence tomography (OCT) has driven the development of a new generation of technologies in OCT, including rapidly tunable, broad bandwidth swept laser sources and spectral domain OCT interferometer topologies. In this work, the operation of a turnkey 1300-nm swept laser source is demonstrated. This source has a fiber ring cavity with a semiconductor optical amplifier gain medium. Intracavity mode selection is achieved with an in-fiber tunable fiber Fabry-Perot filter. A novel optoelectronic technique that allows for even sampling of the swept source OCT signal in \( \vec{k} \) space is described. A differential swept source OCT system is presented, and images of in vivo human cornea and skin are presented. Lastly, the effects of analog-to-digital converter aliasing on image quality in swept source OCT are discussed. © 2005 Society of Photo-Optical Instrumentation Engineers. [DOI: 10.1117/1.1961474]

Keywords: optical coherence tomography; medical imaging; biological imaging; tunable lasers; ophthalmic imaging; skin imaging.

Paper 04137RR received Jul. 20, 2004; revised manuscript received Feb. 28, 2005; accepted for publication Mar. 2, 2005; published online Jul. 15, 2005.

1 Background and Motivation

Optical coherence tomography (OCT) is a noncontact, nondestructive imaging modality that acquires depth-resolved 2- and 3-D images of biological tissue. Since its introduction in 1991, the vast majority of OCT imaging systems have employed the principles of time domain (TD) low coherence interferometry. In low coherence interferometry, an interferometric signal is generated if and only if the pathlength of a sample reflector matches that of the scanning reference reflector. In this approach, spectral information is integrated at a single photodiode, and the interferometric signal for a single sample reflector is given by:

\[
i_{\text{tdoct}}(x) \sim \sqrt{R_R R_S} \int_{0}^{\infty} S(\vec{k}) \cos(2\vec{k}x) d\vec{k},
\]

where \( \vec{k} \) is the optical wavenumber, \( x \) is the pathlength mismatch between the reference and sample reflectors, \( S(\vec{k}) \) is the source spectral density, and \( R_R \) and \( R_S \) are the reference and sample arm reflectivities, respectively. The inefficiency presented by the approach arises because, while all depths of a sample are illuminated, signal is generated in a serial manner by scanning of the reference arm. The magnitude of the inefficiency is on the order of the ratio of the scan depth \( \Delta x_{\text{max}} \) to the coherence length \( l_c \), which is typically \( 10^2 \) to \( 10^3 \). In this context, coherence length refers to the width of the axial coherence function (i.e., axial point spread function).

\[
i_{\text{ssoct}}(\vec{k}) \sim \sqrt{R_R R_S} \int_{\xi}^{\xi+\Delta \vec{k}} S(\vec{k}) \cos(2\vec{k}x) d\vec{k},
\]

\[
q_{\text{sdoct}}(\vec{k}) \sim \sqrt{R_R R_S} \int_{0}^{1/\text{scan}} \int_{\xi}^{\xi+\Delta \vec{k}} S(\vec{k}) \cos(2\vec{k}x) d\vec{k} d\xi.
\]

The swept source OCT signal \( i_{\text{ssoct}}(\vec{k}) \) is a photocurrent integrated over the linewidth of the swept laser source, while the Fourier domain OCT signal \( q_{\text{sdoct}}(\vec{k}) \) is the quantity of photocurrents collected over the duration of an A scan in a charge accumulation detector such as a charge-coupled device or photodiode array.

In 1995, Fercher et al. proposed the use of spectral domain interferometric techniques for the generation of depth-resolved reflectivity profiles (i.e., A scans) in scattering tissues. Two spectral domain techniques were presented. The first technique, which we call Fourier domain (FD) OCT, uses a broadband light source and collects the spectrally resolved interference signal in a detector-arm dispersive spectrometer. The second, which we call swept source (SS) OCT, time encodes spectral information by sweeping a narrow linewidth laser through a broad optical bandwidth. As in time domain OCT, swept source OCT uses a single photodiode detector. The detector responses for swept source and Fourier domain OCT are Eqs. (2) and (3), respectively:

In vivo
compared to time domain OCT remained an open issue. In 1998, Andretzky et al.\(^3\) derived an expression for the sensitivity of Fourier domain OCT that showed its theoretical superior sensitivity to time domain OCT. This expression was later independently rederived and experimentally verified in 1999 by Mitsui\(^4\) and in 2003 by Leitgeb, Hitzenberger, and Fercher,\(^5\) and de Boer.\(^6\) These works, however, overlooked the underlying similarities between swept source and Fourier domain OCT that Fercher et al. recognized. These similarities were exploited by Choma et al.\(^7\) to provide a unified theoretical and experimental sensitivity analysis of all spectral domain OCT techniques that demonstrates the general sensitivity advantage of spectral domain OCT over time domain OCT. This analysis is summarized by two key points. First, the shot noise-limited signal-to-noise ratio (SNR) of an individual spectral domain pixel is given by
\[
\text{SNR}(k_n) = \frac{\rho}{e} P_S(k_n) R_S \Delta t,
\]
where \(k_n\) is the \(n\)th wavenumber measured \((n \in [1, N])\), \(P_S(k_n)\) is the sample power at \(k_n\), \(R_S\) is the sample reflectivity, \(\Delta t\) is the A-scan time, \(\rho\) is the detector responsivity, and \(e\) is the electronic charge. It should be noted that each \(\text{SNR}(k_n)\) is comparable to the standard time domain SNR expression given by Swanson et al.\(^8\) The second key point is that, while signal adds coherently during Fourier transformation from the spectral to the time domain, noise adds incoherently. This leads to a sensitivity advantage of spectral domain OCT on the order of \(N \approx 10^2\) to \(10^3\) (Fig. 1). The SNR of spectral domain OCT is thus:
\[
\text{SNR}_{\text{OCT}} = \frac{\rho}{2e} P_S R_S \Delta t,
\]
where \(P_S\) is the total power incident on the sample.

One of the impediments to the development of robust swept source OCT systems has been the availability of stable, rapidly tunable swept laser sources. Previously swept sources have included grating\(^9\) and prism\(^10\) tuned external cavity lasers and current tuning of a laser diode.\(^11\) Recently, we and others\(^12\) have demonstrated a rapidly tunable swept laser source that employs a fiber laser ring laser configuration with an intracavity wavelength tuning mechanism. This general cavity configuration was first demonstrated in the mid-1990s,\(^13\)–\(^15\) and commercial devices with an erbium-doped fiber ring that use the fiber Fabry Perot tuning mechanism shown in this work have been commercially deployed.\(^16\) We describe the design and operation of a 1300-nm swept source OCT system based on a modified version of this commercially available swept laser source. We demonstrate a novel optoelectronic wavenumber linearization technique that eliminates the need for resampling spectral domain data in software. We show high-quality SS-OCT images of the cornea and anterior segment, and discuss aliasing as a source of image artifact in spectral domain OCT.

2 Swept Source OCT System Design

A schematic of the novel all-fiber swept laser source is shown in Fig. 2. The fiber ring cavity has two important elements. First, the gain medium is a semiconductor optical amplifier (SOA) that achieves population inversion via current pumping. We used a standard polarization-insensitive SOA from InPhenix, Inc. (Livermore, California) (peak wavelength of \(\sim 1300\) nm, small signal gain of \(\sim 20\) dB, a polarization dependent ratio of \(\sim 0.9\) dB, and a saturation power of \(\sim 10\) dBm). Second, addressable wavelength (mode) selection is achieved with a fiber Fabry-Perot tunable filter (FFP-TF), a completely in-fiber device with a <2-dB insertion loss. The FFP-TF is an all-fiber device having a cavity formed by two dielectric mirrors deposited directly onto fiber ends. A thin air-gap within the cavity is used for wavelength tuning and

Fig. 2 Right: Schematic of 1300-nm swept laser source. Addressable cavity mode selection is performed by the fiber Fabry-Perot tunable filter (FFP-TF). A small portion of the laser output is fed to a fixed fiber Fabry Perot interferometer (FFPI), which is monitored by a photodiode (PD). The electrical comb output of this photodiode serves as the pixel clock for the analog-to-digital converter. The large arrows in the cavity represent isolators. The ring laser has a 40% output coupling ratio. SMF: single-mode fiber, SOA: semiconductor optical amplifier. Left: Schematic of FFP-TF. PZT: piezoelectric tuning element.
shorter wavelengths (Fig. 3). This laser achieves a 130-nm sweep range centered at 1300 nm, ~3 mW of output power, and full-width half-maximum bandwidth of ~90 nm. It should be noted that the source output is different on the forward and backward sweep. This is caused by a frequency downshift in the SOA, which is believed to be due to intraband four-wave mixing via carrier-density modulation in the SOA.17–19 Only the higher-intensity forward sweep was used for imaging. We have demonstrated that the sweep rate is scalable into the kilohertz regime.20

To achieve maximal SNR and axial imaging resolution, the digitally sampled SS-OCT data must be evenly spaced in the $k$ domain before Fourier transformation into an A scan. We have implemented an optoelectronic method that eliminates the need for the computationally intense software resampling needed in Fourier domain OCT. A small portion of the laser output is fed into a fixed fiber Fabry Perot interferometer (FFPI) with a free spectral range $\Delta k=372$ radians/m (17.8 GHz). The output of this FFP is monitored with a photodiode. Every time the laser sweeps through $\Delta k$, the photodiode output spikes.21 Since these spikes are evenly spaced in wavenumber, this signal was processed in analog electronics for use of the pixel clock for analog-to-digital conversion of the interferometric signal.

Since $\Delta k$ is broader than the linewidth of the laser, $\Delta k$ determines the maximum scan depth $x_{\text{max}}$

$$x_{\text{max}} = \frac{1}{4\Delta k}.$$  

(6)

The nominal noise equivalent bandwidth (NEB) of the SS-OCT system is thus7

$$\text{NEB} = \frac{2x_{\text{max}}\Delta \lambda}{\lambda_0 f_{\text{ascan}}} = \frac{\Delta \lambda}{2\Delta k f_{\text{ascan}}}.$$  

(7)

where $\lambda_0$ is the center wavelength, $\Delta \lambda$ is the laser sweep bandwidth, $\Delta k$ is $\Delta k$ expressed in wavelength, and $f_{\text{ascan}}$ is the A-scan line rate. Due to nonlinearities in the wavenumber scan, the actual NEB will be greater than that given by Eq. (7). In other words, since the FFPI-based trigger forces even sampling in wavenumber, it necessarily forces uneven sampling in time when the source wavenumber sweep is nonlinear. Since the NEB is inversely proportional to the analog-to-digital sampling interval, nonlinear sampling in time increases the NEB due to the presence of sampling intervals shorter than those assumed in Eqs. (6) and (7). This broadening degrades the system SNR, and is analogous to the broadening observed when a sinusoidal resonant scanner is used in a rapid scanning optical delay line in a time domain OCT system.22

Our system schematic is shown in Fig. 4. The interferometer employs dual-balanced detection,23 which allows for the hardware removal of noninterferometric and autocorrelation terms. Dual-balanced detection extinguishes tens of decibels of autocorrelation and source spectral shape terms. Unfortunately, the fiber optic couplers used by us and others are not achromatic. As a consequence, the coupler splitting ratio is a function of wavelength, which renders complete extinction of autocorrelation and source spectral terms not possible by dual balancing alone. The measured system sensitivity was 119 dB as measured with a −57-dB reflector. The sensitivity is ar-

---

**Fig. 3** Output spectrum of swept laser source. The piezoelectric element in the FFP-TF is driven with a triangle wave, which generates a forward (F) and a backward (B) sweep. During the forward sweep, an increasing voltage ramp is applied to the piezoelectric element, which sweeps the source output from shorter to longer wavelengths. During the backward sweep, a decreasing voltage ramp is applied to the piezoelectric element, which sweeps the source output from longer to shorter wavelengths. This recording was evenly sampled in wavenumber using the pixel trigger mechanism described in Fig. 4 and in Sec. 2 of the text. By using the minimum and maximum source wavelengths measured using an optical spectrum analyzer, the intensity versus time recording was replotted as intensity versus wavelength.
Fig. 4 Schematic of the SS-OCT imaging system. The source sweep function $S(k)$ (k, optical wavenumber) is parametrized by time with the expression $k = k_0 + (dk/dt)$, where $k_0$ is the wavenumber at $t = 0$ and $t$ is taken modulo the total sweep time. The interferometer employs dual-balanced detection, which allows for the hardware removal of noninterferometric and autocorrelation terms. An antialiasing filter mitigates the effect of noninterferometric and autocorrelation terms.

Importantly, the coherence length remained constant throughout the entire 4.2-mm scan depth. The measured coherence length was $9.1 \mu m$ in air and $6.5 \mu m$ in tissue.

3 In Vivo Swept Source OCT Imaging

OCT imaging of the cornea and anterior segment of the eye has important potential applications in glaucoma evaluation and refractive surgery (e.g., LASIK). Compared to ultrasound biomicroscopy, OCT is noncontact and achieves higher spatial resolutions. In terms of imaging the anterior chamber angle with gonioscopy, it has the advantage of being able to visualize structures behind the iris. OCT also has the potential to evaluate in detail the response of the angle structures to light an accommodation, which is an important part of the glaucoma workup. The 1300-nm window is particularly well suited for imaging these anatomical regions because, compared to the 800-nm window, it has deeper penetration into scattering tissue and less penetration into the vitreous. To this end we performed in vivo imaging of the human cornea and anterior segment using slit-lamp biomicroscope adapted optics. Details of this setup are in Patil et al. The image in Fig. 6 contains 300 A scans. Major anatomic features were visualized, including the iris stroma, iris pigment epithelium, ciliary body, angle of anterior chamber, aqueous, corneal stroma, sclera, and scleral spur. The corneal epithelium was visualized in Fig. 7.

OCT imaging of the skin is an emerging technology for subsurface assessment of structure, collagen birefringence, and blood flow. Its resolution is superior to that of ultrasound biomicroscopy. Morphologic changes such as blistering and inflammatory changes such as psoriasis have been observed.
artifact arises because autocorrelation and source spectral shape terms are centered around $x=0$. A third source of ambiguity arises when spectral domain fringes from depths greater than $x_{\text{max}}$ are aliased down to depths less than $x_{\text{max}}$ (Fig. 7). The apparent or aliased depth of an object $x_a$ is given by:

$$x_a = \begin{cases} \mod(x, 2x_{\text{max}}) & \text{if} \mod(x, 2x_{\text{max}}) \leq x_{\text{max}} \\ x_{\text{max}} - \mod(x, 2x_{\text{max}}) & \text{if} \mod(x, 2x_{\text{max}}) > x_{\text{max}} \end{cases}, \quad (8)$$

where $\mod(a, b)$ yields $a$ modulo $b$. This phenomenon has been discussed in Fourier transform spectral interferometry, and a similar effect occurs in magnetic resonance imaging. Aliasing can be mitigated by passing the photodetector output through the noise equivalent bandwidth-limiting antialiasing filter (Fig. 4). Aliasing is a difficult problem to eliminate, because analog filters have finite impulse response transfer functions and, as such, cannot act as perfect brickwall filters with a cutoff frequency corresponding to the system noise equivalent bandwidth dictated by $x_{\text{max}}$. Doubling the scan depth through by resolving complex conjugate ambiguity reduces the likelihood of aliasing. Oversampling and subsequent processing with an infinite impulse response digital filter is another potential solution.

5 Conclusions

We present a swept source OCT system with an all-fiber, turnkey swept laser source. We demonstrate a novel optoelectronic wavenumber linearization technique that yielded a near-transform-limited point spread function performance over the entire scan range, which had not previously been convincingly demonstrated with either Fourier domain or swept source OCT. We obtained excellent in-vivo images of the human cornea and anterior segment, and are currently scaling this system into the kilohertz regime for real-time imaging.

Acknowledgments

This work was supported by National Institutes of Health grant R24-EB000243.

References

7. M. A. Choma, M. V. Sarunic, C. Yang, and J. A. Izatt, “Sensitivity advantage of swept-source and Fourier-domain optical coherence to-

Fig. 7 Effect of aliasing on imaging the cornea and anterior segment. Aliased features are noted by dashed arrows, and A and P denote anterior and posterior, respectively. The irides (I) and lens (L) are anatomically posterior to termination of the 4.2-mm scan, but they are aliased anterior to this termination. Furthermore, aliasing creates mirror images of the irides and lens around the posterior border of the image, consistent with Eq. (8). AC, autocorrelation and spectral shape terms; CE, corneal epithelium; and CS, corneal stroma.

Fig. 8 Swept source OCT image of fingerpad. The bright lines on the left-hand side of the figure are artifact due to detector saturation. The image contains 500 A scans. D, dermis; DEJ, dermo-epidermal junction; and SC, stratum corneum.