Fourier Domain Optical Coherence Tomography for Retinal Imaging with 800-nm Swept Source: Real-time Resampling in k -domain

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In this study, we demonstrated Fourier-domain/swept-source optical coherence tomography (FD/SS-OCT) at a center wavelength of 800 nm for *in vivo* human retinal imaging. A wavelength-swept source was constructed with a semiconductor optical amplifier, a fiber Fabry-Perot tunable filter, isolators, and a fiber coupler in a ring cavity. Our swept source produced a laser output with a tuning range of 42 nm (779 to 821 nm) and an average power of 3.9 mW. The wavelength-swept speed in this configuration with bidirectionality is 2,000 axial scans per second. In addition, we suggested a modified zero-crossing method to achieve equal sample spacing in the wavenumber (k) domain and to increase the image depth range. FD/SS-OCT has a sensitivity of ~ 89.7 dB and an axial resolution of 10.4 μm in air. When a retinal image with 2,000 A-lines/frame is obtained, an acquisition speed of 2.0 fps is achieved.

Keywords : Optical coherence tomography, Swept source, Retina, Ophthalmology

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I. INTRODUCTION

Optical coherence tomography (OCT) has been extensively studied over the past 20 years, as it provides *in vivo* real-time cross-sectional images and three-dimensional volumetric images of biological tissues with a high axial resolution [1]. In ophthalmology, OCT was already commercialized and widely used for non-invasive structural and quantitative imaging of the human retina and anterior segment [2-5]. Recently, Fourier-domain OCT (FD-OCT), which includes a spectral-domain method and a swept-source method, has become a preferred method owing to its superior features of higher acquisition speed and sensitivity [6-10]. Due to the limited number of camera pixels, spectral-domain OCT (FD/SD-OCT) has trade-offs among the axial resolution, imaging depth range, acquisition time, and sensitivity roll-off [11, 12]. FD-OCT based on a swept source (FD/SS-OCT) has overcome these trade-offs. FD/SS-OCT can obtain a low sensitivity roll-off as it realizes a long imaging depth

and reduced fringe washout. In addition, FD/SS-OCT can be used to remove the DC autocorrelation noise by employing a dual-balanced detector. Finally, a faster acquisition speed is possible since the rise time of a photodiode is faster than that of cameras based on charge-coupled device (CCD) and complementary metal-oxide semiconductor (CMOS) chips [13-15].

Although FD/SS-OCT has these advantages, it has been studied, only by a few groups, at 840 nm or 850 nm, which is more useful than FD/SS-OCT at 1310 nm with regard to retina imaging, because the characteristics of optical components in the 800-nm region have limitations. Recently, a swept source in the 1.0 μm region has been actively used for retina imaging because water absorption has a local minimum and the dispersion in water is near zero in the 1.0 μm region [16, 17]. However, a broadband source centered in the 800 nm region rather than in the 1.0 μm region is favorable for obtaining high resolution [18]. Chinn *et al.* [19] have developed an 840-nm swept source

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with a full-width at half-maximum (FWHM) of 20 nm, which utilizes a superluminescent light-emitting diode and a Littrow configuration with a grating and a resonant galvanometer scanner. Lim *et al.* [20, 21], Lee *et al.* [22], and Srinivasan *et al.* [17] introduced an 850-nm swept source with an FWHM over 35 nm employing semiconductor optical amplifiers (SOAs). Lim *et al.* developed a swept source at 850 nm based on a ring cavity including a free-space isolator [20] or a linear cavity with only an optical fiber coupler (fiber loop mirror) [21]. They used the spectral filter comprised of a diffractive grating, achromatic lenses, and a polygon mirror. Lee *et al.* and Srinivasan *et al.* also constructed spectral filters on the basis of Littman and Littrow configurations, respectively [17, 22]. These configurations included a diffractive grating and galvanometer scanner. Although these 850-nm swept sources had a broad spectrum full bandwidth of approximately 60 nm, these did not have the isobestic point, for which the hemoglobin (Hb) and oxygenated hemoglobin (HbO₂) absorption coefficients are equal [23, 24]. However, 800-nm wavelength is very close to the isobestic point. Therefore, the spectroscopic OCT using 800-nm swept source instead of 840-nm or 850-nm swept source can measure functional information such as hemoglobin oxygen saturation level (SO₂) [23] in retina blood vessels. The spectroscopic OCT can provide spectroscopic properties and biochemical properties of biological tissue as well as a cross-sectional image and hemoglobin oxygen saturation level [25, 26].

In this preliminary study, we demonstrate FD/SS-OCT using an 800-nm swept source for *in vivo* human retina imaging. Previous swept sources based on gratings (Littman and Littrow configurations) have a bulky type filter with free space path. Free space path in the swept source sometimes causes large insertion loss, which is a hurdle for achieving a high optical power. Very recently, a fiber Fabry-Perot tunable filter, a fiber-coupled isolator, and an SOA at a center wavelength of 800 nm were commercialized. Therefore, we could easily construct the swept source based on a fiber type ring cavity design without free space path, thus securing low insertion loss. The tuning range of the swept source was from 779-821 nm, and the averaged optical power was approximately 3.9 mW. The wavelength-swept speed in this configuration operating in bidirectional mode is 2,000 axial scans per second. Additionally, in this paper, we suggest a modified zero-crossing method to achieve equal sample spacing in the wavenumber (k) domain. The retina of a healthy volunteer was measured, with an A-line rate of 2.0 kHz, an axial resolution of 10.4 μ m in air (7.6 μ m in tissues), and a sensitivity of 89.7 dB.

II. SWEEP SOURCE SETUP AND CHARACTERIZATION

Figure 1 shows a schematic of a conventional ring-cavity swept source at the center wavelength of 800 nm. We

FIG. 1. Schematic of the wavelength-swept source with a conventional ring cavity at 800 nm. FFP-TF: Fiber Fabry-Perot tunable filter, PC: polarization controller.

constructed a swept source with recently commercialized fiber-based isolators (Opto-Link Corp. Ltd.) and a Fabry-Perot tunable filter (FFP-TF2, Micron Optics). The gain medium is a fiber-coupled SOA (SOA-372-800, Superlum Diodes Ltd.). The SOA has a -3 dB optical gain bandwidth of 20 nm. The SOA was coupled with isolators for a unidirectional ring cavity as shown in Fig. 1. Half of the total optical power fed from the 1×2 fiber coupler is amplified by the SOA at a specific wavelength filtered by the FFP-TF. The filtered light was amplified by the SOA. The other port (50%) of the fiber coupler was used for the output of amplified light. When the FFP-TF was driven using a 1.0 kHz triangular waveform, the swept source achieved an effective bidirectional sweep rate of 2.0 kHz. The SOA was supplied with a current of 160 mA.

Figure 2 (a) is a peak-held spectrum of the swept source measured by an optical spectrum analyzer. This swept source has a center wavelength of 802.3 nm with an average power of 3.5 mW and full scanning range of 42 nm (779 nm to 821 nm). Figure 2 (b) shows the output trace of the swept source in the time domain. Because the optical power and shape of the swept source are dependent on the polarization state in the cavity and on the birefringence of components, the polarization state is controlled by a polarization controller [27]. As shown in Fig. 2 (b), a higher scanning speed of greater than 2.0 kHz causes a significant power difference between the forward scan and the backward scan because of nonlinearity of the gain medium [27-30]. The power of the forward scan is higher than that of the backward scan. Figure 3 (a) and (b) show the point spread functions (PSFs) for various path length differences between the two arms when the wavelength is forward-scanned and backward-scanned, respectively. We measured the maximum imaging depth range (Δz) to be approximately 6.9 mm while sampling points were being increased and the sample mirror was being moved. Therefore, the instantaneous spectral bandwidth could be calculated to be 0.023 nm by the equation of $\Delta z = \lambda_0^2 / 4\delta\lambda$, where λ_0 is the center wavelength and $\delta\lambda$ is an instantaneous spectral bandwidth. The filter line-width of the FFP-TF is 0.02 nm,

FIG. 2. Optical spectra of the swept source. (a) Peak-held spectrum with the center wavelength of 800 nm. (b) The swept source output trace in the time domain when the filter is scanned at 1.0-kHz triangular waveform.

FIG. 3. The point spread functions (PSFs) for various imaging depths when the wavelength is (a) forward-scanned and (b) backward-scanned. The maximum imaging depth range of approximately 6.9 mm is measured. Therefore, an instantaneous spectral linewidth could be calculated to 0.023 nm.

according to the manufacturing company. Therefore, the calculated instantaneous spectral bandwidth was a reasonable spectral line-width value.

III. FD/SS-OCT SETUP AND DATA PROCESSING

We constructed the FD/SS-OCT based on a Michelson interferometer for retinal imaging as shown in Fig. 4 (a). Light from the swept source was incident onto a Mach-Zehnder interferometer module (INT-MZI-850, Thorlabs Inc.). This Mach-Zehnder interferometer module was constructed from a 95:5 optical fiber coupler, three 50:50 optical fiber couplers, and a balanced photodiode as shown in small

box of Fig 4 (a), which is a schematic provided by the manufacturing company. In the Mach-Zehnder interferometer module, 5% of the light from the swept source was used for the interference signal and power monitor. An output port of the Mach-Zehnder interferometer module gave out 95% of the light of the swept source and was coupled with an optical circulator. Light from the Mach-Zehnder interferometer module and the optical circulator was incident on a 2×2 optical fiber coupler and was split into reference and sample arms with 50:50 ratios. The sample arm consists of 2-D galvanometers with silver-coated mirrors (TS8203, Beijing Century Sunny Technology Co., Ltd), an achromatic doublet lens ($f = 30$, Thorlabs Inc.), and a double aspheric ocular lens (40 D, Volk Optical Inc.) as shown in Fig. 4 (a). The 2-D galvanometers were controlled

FIG. 4. Schematic and photo of the FD/SS-OCT system for retinal imaging. (a) schematic of the FD/SS-OCT system. PC: polarization controller. (b) photo of the sample arm in the FD/SS-OCT system. GSD: galvano scanner driver, GS: 2D galvano scanner, CL: collimation lens, L: achromatic doublet lens, OL: ocular lens, CR: chin rest, HH: head holder.

by an analog-output board (PCI-6722, National Instruments Corp.). To correct dispersion differences between the reference and sample arms, we used a dispersion compensation prism pair (AFS-SF10, Thorlabs Inc.) in the reference arm. We used a neutral density (ND) filter in the reference arm to reduce the source intensity noise [31]. Recombined light from the reference and sample arms was incident onto the balanced detector (PDB110A, Thorlabs Inc.) via the circulator and coupler. The interference signals from both the Mach-Zehnder interferometer module and the Michelson interferometer were converted by a high-speed digitizer with a 12-bit, 200-MS/s, and 256-MB onboard memory (PCI-5124, National Instruments Corp.). 2,000 samples were acquired per A-scan and per channel of the digitizer at 2.0 kHz. Figure 4 (b) is a photo of the sample arm at our FD/SS-OCT system for retinal imaging. We used the frame of a keratometer, including joystick, chin rest and head holder. The sample arm was put on the frame of the keratometer as shown in Fig. 4 (b).

Since the FFP-TF was driven by a triangular function, we performed numerical resampling of the raw fringe data by using the zero-crossing method to achieve equal sampling

FIG. 5. Modified zero-crossing method for resampling in the k -domain. (a) zero-crossing points (red triangles) and two points (blue triangles) between the two zero-crossing points in the Mach-Zehnder interference signal, (b) resampled data (red stars) in Michelson interference signal, and (c) point spread functions after Fourier transform without and with resampling.

space in the k -domain. Figure 5 (a) shows the interference signal (black rectangle) from the Mach-Zehnder interferometer module. First, the zero-crossing points (red triangle) in the interference signal were founded by the linear interpolation method, as shown in Fig. 5 (a). The total number of zero-crossing points was 384, and if only these points were used, the imaging depth range would be limited to approximately 1.45 mm. Therefore, we modified the zero-crossing method to increase the number of sampling points and to obtain, consequently, long imaging depth range. After finding the zero-crossing points, two or three points between the two zero positions were selected, and the Michelson interference signal was resampled at each of these points. The intensity values at the resampled positions were obtained by linear interpolation. For example, when the zero positions

(red triangles) and two points (blue triangles) between the two zero positions were selected as shown in Fig. 5 (a), the intensity values (red stars) at the resampled data points of the Michelson interference signal were calculated, as shown in Fig. 5 (b). Therefore, the imaging depth range was increased to approximately 4.3 mm. Finally, the data was zero-filled so that the total number of data points was 2048. When a mirror was used as a sample, Fig. 5 (c) shows point spread functions at the same position after Fourier transformation both without and with resampling.

IV. RESULTS AND DISCUSSION

An averaged optical power of 0.7 mW or less in the sample arm was incident onto a sample in accordance with safe ocular exposure limits set by the American National Standards Institute (ANSI) [32]. We used a -35.3 dB partially reflecting mirror as the sample to measure performance of the FD/SS-OCT system. The dynamic range of our OCT system for forward and backward wavelength-sweep is shown in Fig. 3 (a) and (b), respectively. A decrease of 6 dB was observed within the depth range of approximately 3.0 mm. It was found that the sensitivity of the system, which was determined by adding the sample attenuation constant (35.3 dB), was approximately 89.7 dB at a depth of 0.25 mm. The axial resolution was measured to be 10.4 μm in air, corresponding to 7.6 μm in tissue. Figure 6 shows axial resolution variations as a function of the depth. The measured axial resolution increased beyond 2.0 mm. This degradation indicates that the interpolation process is sensitive to small errors at high fringe frequencies [8, 28]. When a 1,000 A-lines/frame image was obtained, an acquisition speed of 2.0 frames per second (fps) was achieved.

Figure 7 shows cross-sectional retinal images of a healthy volunteer. Figure 7 (a) was obtained without dispersion compensation. Figure 7 (b) is a dispersion compensated retinal image, but resampling was not yet carried out. Neither image

FIG. 6. Axial resolution variation as a function of the depth.

FIG. 7. Retinal images of a healthy volunteer. (a) OCT image with resampling and without dispersion compensation, (b) OCT image without resampling and with dispersion compensation, (c) OCT image with resampling and dispersion compensation using only forward wavelength-sweep, (d) OCT image with resampling and dispersion compensation using only backward wavelength-sweep, (e) OCT image with resampling and dispersion compensation using both wavelength-sweeps; NFL: nerve fiber layer, IPL: inner plexiform layer, INL: inner nuclear layer, OPL: outer plexiform layer, ONL: outer nuclear layer, IS/OS: photoreceptor inner segment/outer segment junction, RPE: retinal pigment epithelium, CH: choroid, BV: blood vessel.

could present clear structural information of the retina. After applying both dispersion compensation and resampling to the original image, detailed structure of the retina was clearly distinguishable. Our swept source is continuously tuned with forward and backward wavelength directions because the swept source is driven by a triangular wave function. We obtained 2,000 lines per frame at both forward and backward wavelength directions and then chose 1,000 lines per frame at each wavelength direction. Figure 7 (c) and (d) were obtained at each swept directions, one is for the forward direction and the other is for the backward direction. We applied a median filter with 3×3 matrix to eliminate speckle noise and selected a region of interest (ROI) with an area of 1,000 (lateral) $\times$ 355 (axial) pixels corresponding to a physical size of approximately 5.0 mm (width) $\times$ 1.5 mm (depth) in air. Figure 7 (e) shows the OCT image using both wavelength-swept directions. Therefore, the pixel size of Fig. 7 (e) is 2,000 (lateral) $\times$ 355 (axial) pixels corresponding to a physical size of approximately 5.0 mm (width) $\times$ 1.5 mm (depth) in air.

FD/SS-OCT for retinal imaging at the 850-nm region has been already studied by Lim *et al.* [24] and Srinivasan *et al.* [18]. This study showed preliminary results for high-resolution and high-speed FD/SS-OCT in ophthalmology. In this study, our swept source still has limitations of the wavelength-tuning range and speed. In the previous papers [24], the wavelength-tuning ranges were 55 nm in the 850-nm region and the FWHMs were broader than our swept source. Although the wavelength-tuning range of our swept source is narrow (42 nm), the axial resolution (10.4 μ m) of our OCT system is lower than that of the previous result (13 μ m) because the center wavelength of our swept source is lower according to the equation: $\Delta z = 0.44 \times \lambda_0^2 / \Delta \lambda$, where $\Delta \lambda$ is the FWHM of the source. Currently, we are improving the wavelength-tuning range and speed of the swept source. Limitation of the wavelength-tuning range can be overcome by using multiple SOAs [33]. Increase of the wavelength-tuning range will also affect the axial resolution of an OCT system. Finally, it is easy to improve the limit of the wavelength-swept speed to hundreds of kilohertz if the cavity length is very short and it is combined with a high-speed filter of kilohertz order such as a polygon scanner or microelectromechanical system (MEMS) scanner [34-36].

V. CONCLUSION

In this study, we devised a wavelength-swept source at a center wavelength of 800 nm and developed an FD/SS-OCT system for retinal imaging. The swept source was constructed using a ring cavity with an SOA, an FFP-TF, isolators, and a 1×2 fiber coupler (50:50 ratios). Our swept source produces laser output with a tuning range of 42 nm (779 to 821 nm) and an average power of 3.9 mW. The wavelength-swept speed in this configuration with

bidirectionality is 2,000 axial scans per second. In addition, the instantaneous spectral bandwidth was calculated to be 0.023 nm. We suggested a modified zero-crossing method to achieve equal sample spacing in the k -domain and to increase an image depth range. Our FD/SS-OCT has a sensitivity of ~ 89.7 dB and an axial resolution of 10.4 μ m in air. We observed a decrease of 6 dB within a depth range of approximately 3.0 mm. When a 1,000 A-lines/frame retinal image was obtained, an acquisition speed of 2.0 fps was achieved. Finally, we were able to obtain *in vivo* a cross-sectional retinal image of a healthy volunteer. Currently, we are developing a wide-band ($>$ FWHM of 100 nm) and high-speed ($>$ 40 kHz) swept source with multi SOAs at 830-nm region.

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