In vivo Fourier-domain full-field OCT of the human retina with 1.5 million A-lines/s

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Received June 8, 2010; revised September 17, 2010; accepted August 17, 2010;
posted September 23, 2010 (Doc. ID 129681); published October 12, 2010

In vivo full-field (FF) optical coherence tomography (OCT) images of human retina are presented by using a rapidly tunable laser source in combination with an ultra-high-speed camera. Fourier-domain FF-OCT provided a way to increase the speed of retinal imaging by parallel acquisition of A-scans. Reduced contrast caused by cross talk was observed only below the retinal pigment epithelium. With a 100 Hz sweep rate, FF-OCT was fast enough to acquire OCT images with acceptable motion artifacts. FF-OCT allows ultrafast retinal imaging, boosting image speed by a lack of moving parts and a considerably higher irradiation power. © 2010 Optical Society of America

OCIS codes: 170.4500, 140.3600, 110.6880.

Since the first demonstration of *in vivo* imaging of the retina [1], optical coherence tomography (OCT) has changed diagnosis in ophthalmology. The increase of imaging speed from 100 A-scans/s to now clinically established 52,000 A-scans/s [2] contributed significantly to the success of OCT by avoiding motion artifacts, enabling volumetric imaging and reducing speckle noise by frame averaging of the OCT data. Today, the fundamental speed limit to retinal imaging is given not by the technology but by the exposure limit to the retina. The limited photon flux, which is allowed to enter the eye, directly links imaging speed and OCT sensitivity, which is in the photon noise limit determined by the number of photons detected in one A-scan. Time-domain OCT (TD-OCT) measured only the photons from one voxel at one time. With the introduction of Fourier-domain OCT (FD-OCT), the acquisition was parallelized with respect to the depth, and at the same sensitivity the acquisition speed could be increased by more than 2 orders of magnitude [3]. Recently, retinal imaging at more than 300,000 A-scans/s was demonstrated at a sensitivity of 89 dB with a complementary metal-oxide-semiconductor (CMOS) camera-based OCT system [4]. For the focused beam, the safety regulations allow for a maximal permissible exposure (MPE) below 1 mW at 850 nm. A further increase of imaging speed without sacrificing sensitivity is only possible by parallel image acquisition in the lateral direction. According to the ANSI standard Z136.1-2007, an extended illumination of the retina increases the MPE by a ratio of the illumination angle α to $\alpha_{\min} = 1.5$ mrad. At the maximal applicable $\alpha =$ 100 mrad, the MPE is 67 times larger, which in principle can be converted into a correspondingly faster imaging.

Full-field OCT (FF-OCT) uses such an extended illumination and a two-dimensional (2D) camera to acquire volumetric OCT images. Impressive images of *ex vivo* samples were published with time-domain FF-OCT [5–8]. Though concepts for *in vivo* FF-OCT were presented [9], the time-domain approach lacks depth parallelization that is needed for fast retina imaging. To benefit from the higher MPE of the eye, FF-OCT has to be combined with FD-OCT. Parallel detection in one dimension with line-field OCT was demonstrated with a combination of a line detector and a swept source [10] and by a parallel spectrometer with a high-speed 2D CMOS camera [11]. Fourier-domain FF-OCT was used for fingerprint analysis [12,13] and *ex vivo* imaging of retina [14,15], but the volume acquisition time of 2 s (76,800 A scans with 256 pixel/s) was too slow for *in vivo* imaging.

In this Letter, we show that recently available CMOS cameras [16] are fast enough for *in vivo* FF-OCT imaging of the retina. A principle disadvantage of FF-OCT, the cross talk between neighboring image points, is usually avoided by spatially incoherent illumination by a conventional light source (halogen lamp, LED) [17], a rapidly moving diffusor, or a multimode fiber [14]. Spatially incoherent illumination increases the complexity of FF-OCT and requires compensation of phase aberrations in the sample arm, usually by a symmetrical setup with high-quality objectives. Here, we show that spatially incoherent illumination is not necessary for imaging the retina with FF-OCT. The requirements for the quality of the optical system including the eye are not higher compared to scanning OCT.

For retinal FF-OCT imaging, a tunable light source (Broadsweeper BS840-01, Superlum), which offers 3 mW of output power and a sweep range of 50 nm around a center wavelength of 850 nm, was coupled by a mono-mode fiber to the input of a Michelson interferometer (Fig. 1). The specified instantaneous linewidth of 0.05 nm corresponded well to a drop of sensitivity (roll off) of 1.5

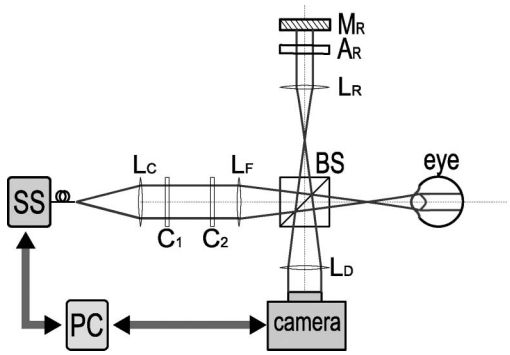

Fig. 1. Schematic drawing of the swept-source full-field setup.

and 6.8 dB in 2 and 3.6 mm depth, respectively. For all imaging optics, achromats (Techspec VIS-NIR achromats, Edmund Optics, USA) were sufficient; the cylindrical lenses were even single elements (Thorlabs, Newton, USA). After collimation by the lens L_C ($f = 50$ mm) and reshaping by the two cylindrical lenses C_1 and C_2 ($f = 75$ mm and $f = -7.7$ mm), the telescope formed by L_f ($f = 125$ mm) and the refractive surfaces of the unaccommodated eye illuminated a 2.6 mm × 0.1 mm rectangular area of retina with a collimated beam of 1.5 mW. A second telescope, which is formed by L_F, the 50:50 beam splitter B_S (BS014, Thorlabs, Newton, USA), and L_R ($f = 25$ mm) projected the collimated rectangular beam via the neutral density filter A_R onto the reference mirror M_R. Both, the retina and the reference mirror were imaged by L_D ($f = 80$ mm) onto the ultrafast CMOS camera (Y4, Redlake/IDT, Tallahassee, Florida, USA). The camera was capable of delivering 5000 images/s at the maximal resolution of 1024 × 1024 pixels, which each had a full-well capacity of approximately 40,000 electrons. The number of lines, which are read out, rather than the exposure time actually limited the frame rate. By reducing the area of interest to 24 lines, frame rates of up to 100,000 fps were achieved, which were just fast enough to prevent image blurring. The lateral image length (640 pixels or 2.5 mm) was a trade-off between field of view and the irradiance on the retina, which determines the sensitivity. Because of the limited radiant flux of the swept laser, a larger field of view would have resulted in an unacceptable low sensitivity. The fading of the signal at the edges is due to the Gaussian beam profile of the retinal illumination. It is a general problem in FF-OCT with spatially coherent illumination. With a pixel size of 14 μm and an overall magnification of 3.6, the image field on the retina was 2.5 mm, corresponding to 6.4°. With 1024 frames per spectral sweep, the equivalent A-scan rate was 1,500,000 lines/s, and volumes of $640 \times 24 \times 512$ voxels were recorded in 10 ms.

After rescaling the spectral data to a linear wavenumber scale and apodization with a Hann window, which reduced the effective width of the spectrum to 25 nm FWHM, a depth resolution of 13 μm in air was determined. For a 4 mm pupil size, a lateral diffraction-limited resolution of 13 μm is expected. The actual resolution may differ because of varying pupil size and additional aberrations. A sensitivity of 72 dB was measured with a sample illumination radiant flux of 92 nW per pixel, which is 11 dB below the quantum-noise-limited sensitivity of 83 dB [Fig. 2(a)]. The difference was attributed to excessive camera noise at the high frame rates and phase noise of the tunable light source. With 10^{-3}, the relative intensity noise was smaller than the 0.5% relative quantum noise of the detector at pixel saturation. Besides statistical noise, a strong fixed pattern noise was also observed, which was especially visible when 50 A scans were averaged [Fig. 2(b)]. The three highest peaks were caused by interference of reflections of the protective camera window. The smaller peaks originate from interference between reflections from optical surfaces in the setup. Because all these peaks are static, they could be subtracted from the image.

With the swept-source FF-OCT setup, retinal images of an emmetropic eye of one coauthor were taken without

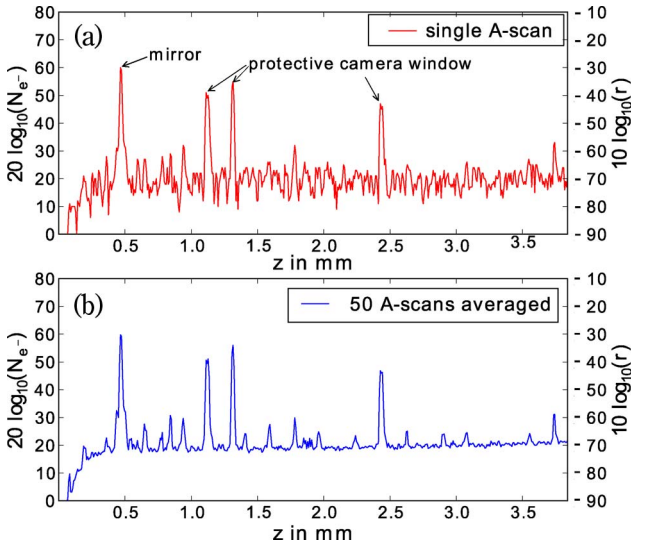

Fig. 2. (Color online) (a) Single A-scan of a mirror imaged through a 1.5 optical density attenuating filter. (b) Averaged image of 50 A-scans. An SNR of 42 dB was achieved. The number of detected photoelectrons N_{e^-} and the corresponding sample reflectivities r are shown on a logarithmic scale.

dilating the pupil. Compliance with the MPE and all relevant safety rules was checked by the local safety officer. At 100 Hz sweeping rate (1024 frames at 100,000 fps), the single A scans from the recorded volume showed the typical layered structure of the retina without severe motion distortion [Fig. 3(a)]. With 70 Hz sweeping rate (70,000 fps), images started to look blurred. With 50 Hz sweeping rate (50,000 fps), it was impossible to record artifact-free images (not shown). The imaging speed was not limited by fringe washout, because the actual integration time is 2–3 orders of magnitude shorter than the A-scan acquisition time. Instead, longitudinal movements introduce a chirp to the spectral interference signal, which distort image geometry and reduce the depth resolution. From the chirp in our data, we expect peak motion velocities to be significantly below 4 mm/s, which would correspond to 40 μm linear motion. The average effect of the chirp was less obvious, and this work shows that motion artifacts are not directly visible in the retinal images when taken at 100 Hz.

At 100,000 fps, the quality of the *in vivo* retina OCT images suffered mainly from the low sensitivity of 72 dB, which was caused by the camera noise and the low photon flux per pixel. Compared to Potsaid *et al.* [4], retinal imaging was 5 times faster with only twice the radiant flux but showed comparable quality. The retinal exposure was limited by the output of the light source and not by the retinal damage threshold. To simulate a measurement with 36 mW, which would be 72% of the MPE, all 24 A scans of one volume were averaged [Fig. 3(b)]. Increased SNR and speckle averaging improved the image quality considerably.

Astonishingly, in spite of expected cross talk or interference between reflections from the different optical surfaces of the eye, the layered retinal structure above the retinal pigment epithelium (RPE) is imaged with good quality. Compared to OCT images with the Heidelberg Engineering Spectralis (images not shown), the averaged retina image in Fig. 3(b) is slightly blurred and very fine

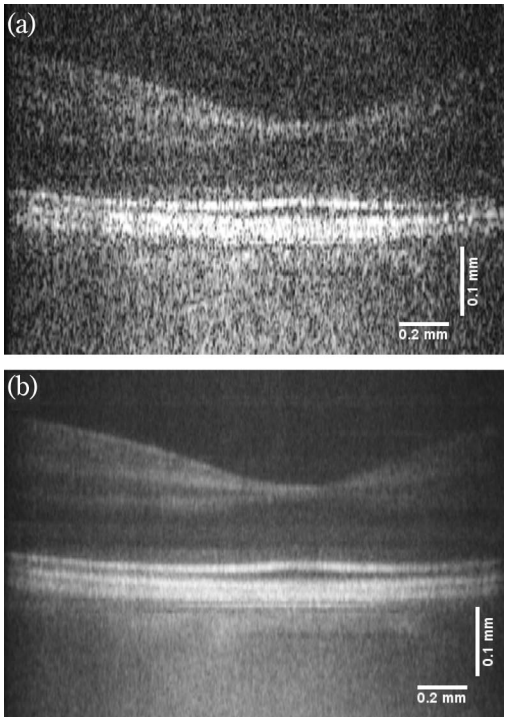

Fig. 3. (a) Single B-scan of human retina recorded with 1,500,000 A-scans/s at 1.5 mW radiant flux. (b) Averaged image of 24 A-scans of the imaged volume.

lateral details are missing. This may be caused by a combination of eye movements, aberrations, and cross talk. Below the RPE, a more-or-less structureless OCT signal appears, which masks the image of the choroid owing to the strong scattering in the RPE. Incoherent illumination of the retina is expected to decrease this stray light background but will considerably increase the technical complexity of the interferometer.

Swept-source FF-OCT combines a simple setup without any moving parts with a high imaging speed by taking advantage of spatially parallel image acquisition, which allows for a significantly higher retinal exposure. At a sweep rate of 100 Hz and 1.5 mW radiant flux, the *in vivo* OCT images of the retina were limited in quality mainly by the relatively low radiant flux per pixel and by camera noise. With our setup, the sensitivity of FF-OCT is still too low to compete with commercial OCT devices, which usually provide more than 95 dB. However, faster CMOS cameras with quantum-noise-limited performance, balanced detection, as demonstrated by Sacchet *et al.* [9], and an increase of the radiant flux by a factor of nearly 35, which fully exploits the MPE for an extended light source, could raise the sensitivity to 96 dB. Besides these technical challenges, which put FF-OCT still far from being competitive with high-performance clinical scanning OCT systems, it also has principle disadvantages with regard to cross talk and uniform image quality. However, FF-OCT is one way to overcome speed limitation in retinal OCT imaging set by the scanner and the limited retinal exposure. In addition, the simple setup without fibers or moving parts could be of advantage for polarization-sensitive OCT, Doppler, or profilometric measurements.

This work was partly funded by the European Union project FUNOCT (FP7 HEALTH, contract no. 201880). We thank Prof. F. Ùberle for supervising the Master Thesis of G. F. and Dierck Hillmann for valuable discussions.

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