Published in final edited form as:

Opt Lett. 2010 December 1; 35(23): 3913–3915.

Sample motion-insensitive, full-range, complex, spectral-domain optical-coherence tomography

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Abstract

We present a full-range, complex, spectral-domain optical-coherence-tomography (SD-OCT) system that is based on a double-beam scanning approach. The sample beams of two identical SD-OCT setups are combined collinearly by a bulk optic beam splitter before illuminating the object. The required phase shift for the complex signal reconstruction comes from the phase difference between both interferometers. Because of the double-beam scanning approach, our system is completely insensitive to sample motion. To demonstrate the performance of our setup, we present images of the human optic nerve head *in vivo* and of a human tooth.

Spectral-domain optical-coherence tomography (SD-OCT) has shown to be a powerful technique for imaging biological samples [1]. The depth-resolved information about the object is encoded in the cross-spectral density function, which is measured with a spectrometer. One drawback of this method is that the detected spectrum is a real valued function, and hence its Fourier transform is Hermitian. Therefore, the reconstructed images are symmetrical with respect to the zero path-delay difference. This is of minor importance for measuring thin objects (e.g., the fovea region of the retina), because the reference arm can be shifted to a position where the structure and mirror term are not overlapping. But if one wants to measure objects with larger depth extensions (e.g., anterior segment of the eye or optic nerve head) where the whole measurement range is needed, the mirror term has to be removed in order to obtain unambiguous images. SD-OCT systems also suffer from a sensitivity decrease with distance from the zero delay position. Therefore, a differentiation between negative and positive path-length differences would be highly desirable, because this would allow us to place the object within the most sensitive measurement region.

It was shown that this problem could be overcome by measuring the phase of the spectral density function, thus providing access to the complex interferogram [2]. Various different full-range complex SD-OCT (FRC SD-OCT) systems were reported during the past few years. These methods employ piezo-driven mirrors [3-5], electro-/acousto-optic modulators [6,7], off-pivot-point illumination of the galvanometer scanner [8-10], or polarization-based optical demodulation [11] to introduce a phase shift, required to reconstruct the complex-valued signal. The main drawback of most of these methods is that they are sensitive to sample motion. If the object moves during the scan, an undesired phase shift is introduced that corrupts the reconstruction of the complex-valued signal.

In this Letter, we present an approach to overcome this problem. A schematic diagram of our setup is shown in Fig. 1. It consists of two identical fiber-based SD-OCT systems. For

light sources, we use two superluminescent diodes with a center wavelength of 840 nm and an FWHM of 50 nm. Each spectrometer unit consists of a diffraction grating with 1200 lines/mm, a 200 mm focal length lens, and a line-scan camera with 2048 pixels. This setting provides a total depth range of 6.2 mm (2×3.1 mm). Light in the reference arm is attenuated by a variable neutral density filter to drive the CCD cameras shortly below the saturation limit. A prism pair in each reference arm is used to compensate the dispersion mismatch with the sample arm. In the object arm, a bulk optic 50:50 beam splitter combines both sample arm beams collinearly before the galvanometer scanner. Each SD-OCT system provided a probing power of 0.35 mW, and the integration time of the cameras was set to 70 μ s. With these settings, we measured a total sensitivity of 98 dB and an attenuation of 12 dB at $z = \pm 2.3$ mm. Assuming that both SD-OCT systems are properly aligned and that both beams in the sample arm are truly collinear, we record the same spectral density function at each line-scan camera. The phase difference between both channels, which is required to reconstruct the complex scattered field, comes from the phase difference between both interferometers. This phase difference can be set by adjusting the reference arm length properly. Note that there is no cross talk between the two sample arm beams, because they are provided by two individual super luminescent diodes (SLDs).

The reconstruction algorithm is based on the work by Leitgeb *et al.* [4] and is outlined below: assuming that the phase difference between both interferometers is 90° , we record two spectrograms $I(\nu)$ and $I(\nu, \Delta\phi = 90^\circ)$. Both signals real valued are functions, and, therefore, an inverse Fourier transformation contains the structure and mirror terms as well as the dc and autocorrelation terms [6]:

$$\text{FT}^{-1}\{I(\nu)\} = \Gamma_{rr}(\tau) + \sum_n \Gamma_{nn}(\tau) + \sum_{n \neq m} \{\Gamma[\tau + (\tau_m - \tau_n)] + \Gamma[\tau - (\tau_m - \tau_n)]\} + \sum_n \{\Gamma[\tau + (\tau_r - \tau_n)] + \Gamma[\tau - (\tau_r - \tau_n)]\}. \quad (1)$$

In this equation, Γ denotes the first-order electric field correlation function, the indices m and n run over different reflecting object layers, r denotes the reflectance from the reference arm, and $\tau_{n,m,r}$ are the traveling times of the beam. The first two terms in this equation correspond to the dc peak, the second term represents the autocorrelation terms, and the last term contains the structure and mirror terms.

As a first postprocessing step, we subtract the mean spectrum from each spectrogram (corresponding to an individual A scan) to remove dc and autocorrelation terms [12]. Then we apply a simple algorithm that corrects for small phase and amplitude errors between both channels. We calculate the inverse Fourier transform of $I(\nu)$ and $I(\nu, \Delta\phi = 90^\circ)$ and subtract one from the other to display phase and amplitude differences between them. This information is used to correct $\text{FT}^{-1}\{I(\nu)\}$ such that the phase difference between both images is exactly 90° and to eliminate any amplitude differences. Then we backtransform both signals via a Fourier transform to obtain two corrected real valued spectra $I_c(\nu)$ and $I_c(\nu, \Delta\phi = 90^\circ)$. These spectra can be seen as quadrature components from which the complex spectrum can be reconstructed:

$$\tilde{I}_c(\nu) = I_c(\nu) + iI_c(\nu, \Delta\phi = 90^\circ). \quad (2)$$

An inverse Fourier transform of this new complex spectral density function yields the true object structure without the mirror term:

$$S_c(\tau) = \text{FT}^{-1}\{\tilde{I}_c(\nu)\} = 2\sum_n \Gamma[\tau + (\tau_r - \tau_n)]. \quad (3)$$

To show the performance of our system, we present images of the human nerve head *in vivo*. Figure 2(a) shows the reconstructed depth profile from one of the two cameras. In this image, the dc and autocorrelation terms are already removed, but the mirror term is still present. Figure 2(c) shows signal $S_c(\tau)$ where the mirror term is nicely removed. Extinction ratios of ~30 dB are observed. To present the efficiency of our phase and amplitude correction algorithm, we show one image [Fig. 2(b)] where this reconstruction step was left out. Residual mirror terms are clearly visible. This shows that our system is rather immune against small phase and amplitude errors. But if larger differences, e.g., phase errors $>90^\circ$ occur, this correction algorithm fails, because then the structure and mirror term are exchanged.

One important feature of our dual-beam FRC SD-OCT system is its ability to achieve full-range imaging regardless of sample motion. Assuming that both sample arm beams are collinear, any additional phase shift due to object movement is automatically compensated for, because both beams undergo the same phase shift. To demonstrate this fact, we show OCT B scans of a human tooth that was mounted on a motorized stage. During the measurement, the tooth was moved with a constant velocity of 1.5 ± 0.1 mm/s in the beam direction. In addition to that, we also offset both scanning beams from the pivot point of the galvanometer scanner to compare our FRC SD-OCT system with the one described in [8]. Figure 3(a) shows an image where no complex signal reconstruction algorithm was applied, and therefore the mirror term is still present. (It should be noted that the lower image quality of Fig. 3 is caused by phase wash-out due to the sample motion.) In Fig. 3(b), one can see the resulting image from the complex reconstruction algorithm described in [8]. Because of the additional phase shift, which is introduced by the sample motion, this reconstruction algorithm fails. Figure 3(c) shows the resulting image from our new dual-beam reconstruction algorithm. One can see that the sample motion does not influence the suppression of the mirror term at all.

The main disadvantage of our current setup is the required phase stability between both SD-OCT systems. Because of thermal drifts and vibrations, the phase difference is constantly varying. This is of minor importance for a single B scan, because we observed no fast phase fluctuations between both interferometers (~ms range). But it prohibits full 3D scans, which require 2–3 s of measurement time. On this time scale, larger phase drifts ($>90^\circ$) might occur, which causes the mirror and structure term to be exchanged, leading to an incorrect image reconstruction. One could overcome this problem by inserting an active phase stabilization mechanism.

Acknowledgments

The authors thank H. Sattmann and C. Wöfl for technical assistance and the Austrian Science Fund (grant P19624-B02) for financial support.

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Fig. 1.

Schematic diagram of the dual-beam FRC SD-OCT setup: FBS, 50:50 fiber beam splitter; FC, fiber coupler; VDF, variable neutral-density filter; DCP, dispersion-compensating prism pair; M, mirror; DG, diffraction grating; L, lens; LSC, line-scan CCD camera; BS, bulk optic 50:50 beam splitter; GS, galvanometer scanner; S, sample.

Fig. 2. OCT B-scan images of the human nerve head *in vivo* displayed on a logarithmic gray scale. (a) Image obtained by an inverse Fourier transform over one real valued spectral data from one camera. (b) Image obtained from reconstructed complex spectrum without phase and amplitude correction algorithm. (c) Image obtained from reconstructed complex spectrum with correction algorithm.

Fig. 3. OCT B-scan images of a human tooth that was moving at constant velocity of 1.5 ± 0.1 mm/s along the beam direction. All three images are reconstructed from the same dataset and displayed on a logarithmic gray scale. (a) Image obtained by an inverse Fourier transform of one real valued spectral dataset from one camera. (b) Image after complex signal reconstruction algorithm described in [8]. (c) Image obtained with the new dual-beam scanning method, which shows no sensitivity to sample motion.