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Simultaneous swept source optical coherence tomography of the anterior segment and retina using coherence revival

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Abstract

We report on an implementation of coherence revival-based heterodyne swept source optical coherence tomography that is capable of simultaneously imaging the anterior and posterior eye. A polarization-encoded sample arm was used to efficiently focus orthogonal polarizations on the anterior segment and retina. Depth encoding was achieved using coherence revival, which allows for multiple depths within a sample to be simultaneously imaged and frequency encoded by carefully controlling the optical pathlength of each sample path. This design is a significant step toward whole-eye optical coherence tomography (OCT), which would enable customized ray-traced modeling of patient eyes to improve refractive surgical interventions and eliminate optical artifacts in retinal OCT diagnostics. We demonstrated the feasibility of this system for *in vivo* imaging by simultaneously acquiring images of the anterior segments and retinas in healthy human volunteers.

Keywords

110.4500; 110.4190; 170.4460

Optical coherence tomography (OCT) provides micrometer scale resolution of tissue structures, and has become the standard of care for the diagnosis of many human retinal pathologies. In addition to acting as a powerful image-based diagnostic tool in the posterior eye, OCT also has the potential to provide quantitative biometry of the ocular anterior segment, including measurements of corneal refractive power, corneal thickness, anterior chamber depth, crystalline lens optical power, and crystalline lens thickness [1-3]. Current methods for evaluation of these parameters are limited to measurement along a single axis and do not accurately account for off-axis variations and aberrations. With the ability to rapidly acquire densely sampled 2D images and 3D volumes, OCT offers the potential to perform substantially improved characterization of the refractive properties of the entire eye. This would enable improvements in customized modeling of patient eyes. Such models would enhance intraocular lens customization and refractive surgical interventions, and also

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facilitate the correction of optical distortions due to the cornea and crystalline lens in conventional retinal OCT diagnostics.

As OCT develops into an important tool for anterior segment characterization, the ability to use a single system for both anterior and posterior imaging becomes a desirable feature for clinical systems. Several commercial ophthalmic OCT systems are capable of switching between anterior segment and retinal imaging modes, allowing a single system to function in both capacities. However, switching between modes requires the addition or removal of optics from the sample path and adjustment to the length of the reference path. Also, as these systems are designed primarily for retinal imaging, anterior segment imaging performance (particularly axial and lateral field of view) is typically suboptimal.

In this work, we describe a novel OCT system that is capable of imaging both the retina and the anterior segment simultaneously. This is achieved by combining coherence revival-based swept source OCT (SSOCT) with a polarization-encoded sample arm. To demonstrate the feasibility of this technique for *in vivo* ophthalmic imaging, the system was used to simultaneously image the anterior segments and retinas of healthy human volunteers.

Coherence revival-based SSOCT [4] is a recently developed technique for resolving the complex conjugate ambiguity in SSOCT. Here, we describe an extension of the technique presented in [4] that enables simultaneous imaging of two different regions of the eye. This was achieved through the use of a dual-channel sample arm that simultaneously scanned the anterior segment and retina. The path through the sample arm that imaged the retina (the “retinal path”) was matched to the reference arm pathlength, encoding the retinal image at the baseband for conventional SSOCT imaging. The “anterior segment path” was offset from the reference arm path-length by a distance equal to the laser cavity length, which encoded the anterior segment interferogram with a carrier frequency. As a result, after inverse Fourier transformation of the acquired spectral interferogram, the retinal image appeared near DC while the anterior segment image appeared shifted in depth. This precluded overlap of the two images while also resolving the complex conjugate artifact for the anterior segment image.

A schematic of the dual-channel sample arm is shown in Fig. 1. The incoming polarization was controlled by the half-wave plate before reaching the first polarizing beam splitter (PBS). Light emerging from the single-mode fiber was partially polarized, such that rotation of the wave plate could direct as much as 83% of the incident light to either path. The transmitted polarization experienced a fixed delay, whereas the reflected polarization experienced a longer and variable delay that could be adjusted via a translation stage (dashed box in Fig. 1). The two polarizations were then recombined at the second PBS and scanned by a pair of orthogonal galvanometer mirrors. Reflections at the galvanometer mirrors resulted in a rotation of the polarization, such that the *P* and *S* polarizations were switched. When the scanned beams reached the third PBS, the polarization state that had initially been reflected was then transmitted, and vice versa. Following this second split, each path contained carefully designed optics optimized for either anterior segment or retinal imaging, allowing for the acquisition of images of both regions of the subject’s eye with no

adjustment of the sample arm optics. The two paths were then recombined at the fourth PBS and directed to the patient eye.

Sample arm optics were optimized in Zemax to provide the optimal tradeoff between resolution, aberration, and depth of focus for each imaging depth using off-the-shelf achromatic doublets. The retinal imaging optics were optimized using an eye model with a gradient index lens [5], and designed to scan a 2.6 mm beam incident on the cornea, pivoting through the pupil at about a 17.4 degree scan range. The anterior segment path was designed to have a lateral resolution of 30 μm (defined as the FWHM of the point spread function), and a 15.1 mm lateral scan range. This lower lateral resolution allowed for a much longer depth of field to permit imaging through the entire anterior chamber. The axial resolution for both paths was < 10 μm , limited by the spectral shaping window applied in processing.

Key features of the combined polarization and depth encoding schemes used in this system are optimal optical power conservation and crosstalk rejection. The two sets of PBS cubes effectively created two polarization channels. Neglecting the finite extinction ratios of the PBS cubes, all of the light whose polarization was not modified in the sample returned in the appropriate polarization channel, thereby limiting the sensitivity loss in each channel for dual-channel imaging to a single factor of -3 dB. Returning light whose polarization was modified either by depolarization or birefringence in the sample (and thus contaminated the wrong polarization channel) was also depth encoded and thus lost to imaging, but did not degrade the other image.

The SSOCT system topology is also shown in Fig. 1. An external cavity tunable laser (Axsun Technologies) operating at a 100 kHz sweep rate, output power of 22 mW, central wavelength of 1040 nm, and bandwidth of 100 nm was used as the source. A spectrally balanced Michelson fiber interferometer design [6] was used. While this interferometer design contains an inherent 3 dB inefficiency, it provides near-ideal spectrally balanced detection, which significantly reduces relative intensity noise (RIN) from the source. The power incident on the patient eye was 900 μW per channel, for a total of 1.8 mW. This is in compliance with the most stringent interpretation of the ANSI Z136.1 standard. Light returning from the interferometer was detected on a 1 GHz InGaAs balanced receiver (WL-BPD1GA, Wieserlabs) and digitized at 1 GS/s on an 8 bit (6.7-ENOB), 500 MHz bandwidth digitizer (ATS9870, Alazar Tech). Data acquisition and processing were performed in LabVIEW and C. Separate dispersion compensation algorithms were applied for each depth, which is a requirement of the coherence revival technique [4].

For each channel of the sample arm, power efficiency, peak sensitivity, and sensitivity falloff profiles were measured. Power efficiency measurements were made by placing a mirror in the sample arm and measuring the optical power returning in the power meter channel as a fraction of the power incident on the sample. The retinal channel power efficiency was measured to be 40%, whereas the anterior segment channel efficiency was measured to be 38%.

Peak sensitivity and sensitivity falloff profiles were measured by placing a -48 dB calibrated reflector in the sample arm and translating the reference mirror. Sensitivity was then

computed as the ratio of the peak of the A scan to the standard deviation of the noise floor measured with the sample arm blocked. The peak sensitivity was 96 dB for retinal imaging and 95 dB for anterior segment imaging; the 1 dB difference is a consequence of the coherence revival falloff envelope [4]. The theoretical, shot-noise limited sensitivity was computed, as described in [7], to be 102 dB for each channel; this value includes the inherent 3 dB inefficiency of the spectrally balanced interferometer. The discrepancy between the theoretical and measured sensitivities can be partially explained by optical losses (4 dB). The remaining 2 dB discrepancy was likely due to digitization noise and residual RIN not rejected by spectral balancing.

Sensitivity falloff profiles for each imaging channel are shown in Fig. 2. For the retinal channel, we observed an imaging range of 5.2 mm in air (3.9 mm in tissue), defined as the region over which the sensitivity loss measured from the zero pathlength difference position was less than 6 dB. For anterior segment imaging, the peak sensitivity position appeared at approximately 6 mm in depth, and the anterior segment imaging region spanned approximately 8.5 mm (6.4 mm in tissue), from 1.5 mm to 10 mm. A full doubling of the 5.2 mm imaging depth was not achieved due to limited electronic bandwidth of the digitizer. Thus, there existed a 1.5 mm region in which the retina could be imaged without limiting the imaging range of the anterior segment. While it is possible for faint anterior segment structures to appear in this 1.5 mm region, this was avoided by managing the orientation of each image, as described below. This 1.5 mm retinal imaging range is comparable to the imaging depth of clinical spectral domain OCT systems, and could be extended at the expense of anterior segment imaging range if necessary.

To demonstrate feasibility *in vivo*, images were acquired from three healthy human volunteers. Dual-depth B-scans were acquired at 50 Hz for 2000(lateral) $\times$ 2304(axial) pixel images. Figure 3A shows a typical B-scan as observed during acquisition. The retina appears “above” the anterior segment, and both images are upright. This orientation was chosen, as it precludes the possibility of crosstalk between deeper structures in the anterior segment and the retinal image. Figures 3B and 3C show images from the same data set after further processing, including dispersion compensation for the anterior segment image, cropping, scaling, and averaging. The anterior segment image appears slightly angled because the optical axis of the system was aligned with the volunteer’s visual axis, and thus offset from the optical axis of the eye by a few degrees. As a result, the fovea was centered in the retinal image.

We have reported on an SSOC system design that enables simultaneous imaging of the retina and anterior segment. This design employs polarization encoding and coherence revival to discriminate each imaging depth, and allows for the optical design of each sample path to be optimized independently. While coherence revival-based frequency encoding is only applicable to certain sources, the polarization-encoded sample arm design is applicable to any extended depth SSOC system, provided that the relative delay between the two sample paths is adjusted accordingly. This design is a significant step towards whole-eye OCT, the realization of which would enable customized ray-traced modeling of patient eyes. Such customized models have the potential to improve refractive surgical interventions and to eliminate optical artifacts in retinal OCT diagnostics.

Acknowledgments

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Fig. 1.

(Color online) SSOCT system schematic. BR; balanced receiver; PM, power meter; UP, unused ports; PC, polarization controller. Sample arm: red and blue lines depict the retinal and anterior segment imaging paths, respectively. Overlapping paths are shown in purple. L, lens; HWP, half-wave plate; PBS, polarizing beam splitter; FM, fold mirror; G, galvanometers.

Fig. 2.
(Color online) Sensitivity falloff profiles for the retinal (top) and anterior segment (bottom) imaging channels. Red horizontal lines are drawn at -6 dB sensitivity from the peak.

Fig. 3.

(Color online) Simultaneously acquired anterior segment and retinal images. A, single frame as acquired before cropping, consisting of $2000(\text{lateral}) \times 2304(\text{axial})$ samples, acquired in 20 ms. ZPD, zero pathlength difference position; +1 CL, cavity length offset position. B and C, separated anterior segment and retinal images, averaged over 5 frames. Scale bars are 1 mm, $1 \text{ mm} \times 1 \text{ mm}$, and $250 \mu\text{m} \times 1^\circ$ in A, B, and C, respectively. The dynamic ranges of the image data in B and C were 56 and 40 dB, respectively.