Adaptive optics optical coherence tomography for retina imaging

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When optical coherence tomography (OCT) is used for human retina imaging, its transverse resolution is limited by the aberrations of human eyes. To overcome this disadvantage, a high resolution imaging system for living human retina, which consists of a time domain OCT system and a 37-elements adaptive optics (AO) system, has been developed. The AO closed loop rate is 20 frames per second, and the OCT has a 6.7-μm axial resolution. In this paper, this system is introduced and the high resolution imaging results for retina are presented.

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Optical coherence tomography (OCT)[1,2] is similar to ultrasound B mode imaging, however it uses light instead of sound as the imaging source. It has a high resolution of $1-15$ μm, which is one or two orders of magnitude higher than that of the conventional ultrasound technique. OCT can also be used to perform real-time cross-sectional tomography imaging. The unique features of this technology enable a broad range of clinical applications.

Human eye is not a perfect optical system, and it has wave front aberrations[2] which deteriorate the transverse resolution of OCT. The aberrations are different from person to person and change with time, it is therefore difficult to correct the aberrations by a static way. Adaptive optics (AO) can correct the aberration in real time but traditionally it has a bulky size, complicated configuration and high cost, and it is mainly used in astronomical telescope. With the development of low-cost components, AO has been applied in ophthalmic imaging[3−7]. Recently, AO combined with a spectral domain OCT was successfully used in the retinal cone mosaic imaging[6,7].

We have established a time domain AO-OCT system for retina imaging. The optical arrangement of the AO-OCT system is shown in Fig. 1. The configuration of OCT is in the broken-line rectangle. Light from a superluminescent diode (SLD) ($\lambda = 840$ nm, $\Delta\lambda = 49$ nm) is equally split by a fiber coupler into reference and sample arms. The light of sample arm passes through the AO system and is reflected back to the fiber coupler, then it interferes with the light of reference arm which is reflected by a rapid scanning optical delay line (RSOD). The interference signal is measured by a detector and then processed by a computer. In order to compensate the polarization mode dispersion of the system, a fiber polarizer control is applied in the sample path to control the polarization. The carrier frequency can be supplied by RSOD with the phase modulator.

The AO system consists of a 37-element small piezo material lead zirconate titanate (PZT) deformable mirror, a 16×16 array Shack-Hartmann wave front sensor, a beam ($\lambda = 780$ nm), and an x-y scanning system. The beam passes through the whole optical system and is focused on the retina, thereby it experiences the wave front aberrations of the eyes. The wave front sensor detects the back-reflected light and measures the displacements of light centroids in all sub-apertures. The wave front slope could be gotten from the centroid displacements. The control signal calculated from the wave front slope is amplified by a voltage amplifier, then used to control the deformable mirror to correct the wave front aberrations of eyes. The AO closed loop rate is 20 frames/s, which leads to achieve the diffraction-limited resolution in real time.

The benefit of RSOD is that it can compensate the material dispersion of the eye, and combined with the polarization controller, the whole system's dispersion

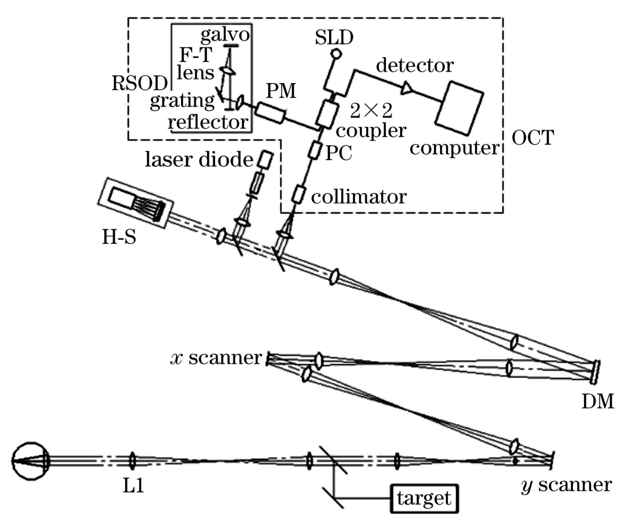

Fig. 1. Schematic of AO-OCT system. PM: phase modulator, PC: polarization controller, H-S: Shack-Hartmann wave front sensor, DM: small PZT deformable mirror, L1: focusing lens.

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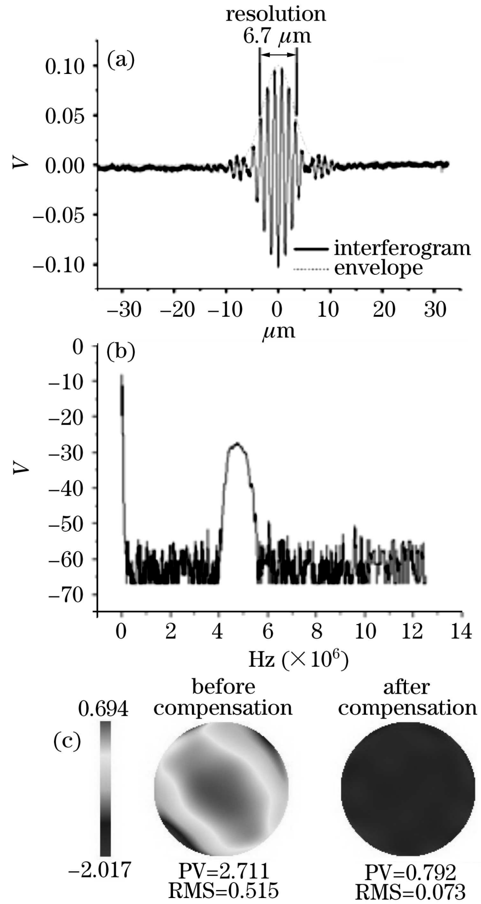

Fig. 2. (a) Interference signal of OCT; (b) FFT of interference signal; (c) aberration before and after compensation.

including material dispersion and polarization mode dispersion could be compensate primely. The system interference signal is shown in Fig. 2(a). The resolution of the OCT is about 6.7 μm, which approaches the ideal resolution of 6.23 μm. The fast Fourier transform (FFT) of the interference signal shows that the carrier frequency is 500 kHz, as shown in Fig. 2(b).

We carry out experiment with a pig eye, and it is fixed by an adjustable mount. Because the pig eye lose focusing ability, so the L1 function as a focusing lens makes the incidence light be focused onto the retina, and the incidence beam diameter is 6 mm with 325-μW power. Figure 2(c) shows the aberration before and after compensation. While the pig eye's aberration is uncorrected, the whole system cannot obtain the diffraction limited resolution. But after correcting the aberration by AO system, the wavefront root-mean-square (RMS) is 0.073λ, so we could guarantee the whole AO-OCT system to achieve ideal transverse resolution.

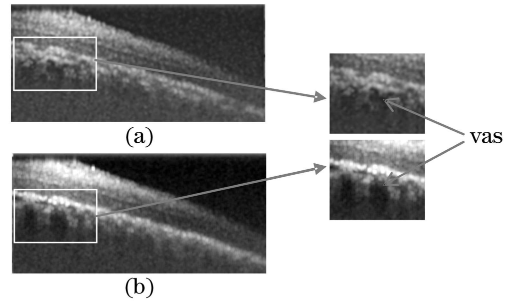

Fig. 3. Retina image (a) without and (b) with AO.

Figure 3 illustrates the effect of aberration correction in OCT, depicting cross-sectional AO-OCT tomograms of a pig eye in the retina across a transverse line of 1.5 mm (500 A-scans) for the uncorrected (Fig. 3(a)) case, as well as corrected (Fig. 3(b)) case. Comparing the same region (white rectangle) of these two images, the vas of retinal photoreceptor (RPE) layer in Fig. 3(b) is visible while it is blurry in Fig. 3(a). So it is obviously that the AO system could enhance the OCT transversal resolution effectively.

In conclusion, we set up a time domain AO-OCT system, which consists of a 37-element small PZT deformable mirror, a 16 × 16 array Hartman-Shack wave front sensor, and a time domain OCT system. The AO closed loop rate is 20 frames/s, so that it can achieve the diffraction-limited resolution in real time. The current experimental results show that the AO can enhance the transverse resolution of OCT effectively.

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