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Sapphire ball lens-based fiber probe for common-path optical coherence tomography and its applications in corneal and retinal imaging

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Abstract

We describe a novel common-path swept source optical coherence tomography fiber probe design using a sapphire ball lens for cross-sectional imaging and sensing for retina vitrectomy surgery. The high refractive index ($n = 1.75$) of the sapphire ball lens improves the focusing power and enables the probe to operate in the intraocular space. The highly precise spherical shape of the sapphire lens also reduces astigmatism and coma compared to fused non-spherical ball lenses. A theoretical sensitivity model for CP-OCT was developed to assess its optimal performance based on an unbalanced photodetector configuration. Two probe designs—with working distances (WD) 415 μm & 1221 μm and lateral resolution 11 μm & 18 μm —were implemented with sensitivity up to 88 dB, which is significantly higher than previously reported CP-OCT probes. We assessed the performances of the fiber probes by cross-sectional imaging a bovine cornea and retina in air and in vitreous gel with a 1310-nm swept source OCT system. To the best of our knowledge, this is the first demonstration of sapphire ball lens-based common-path OCT probes directly inserted into the vitreous gel of a bovine eyeball for ocular imaging with a sensitivity approaching the theoretical limitation of CP-OCT.

Optical coherence tomography is being widely used for non-destructive, cross-sectional imaging of biological tissues [1–6]. A single-mode fiber can be lensed with state-of-the-art micro-optics to form an imaging beam with a spot size around 20 μm ~50 μm in gastrointestinal endoscopy, coronary artery imaging, and needle-based Doppler OCT. The commonly used lensing components in fiber-optic microprobes are gradient-index (GRIN) lenses [7, 8], drum lenses [9], fiber fused ball lenses [10], and special liquid-forming ball lenses. For retina vitrectomy surgery, a handheld compact forward sensing and imaging probe should be attached to surgical tool tips to detect the distance between the tool tips and critical areas of the retina, so they can avoid scratching healthy retina surfaces [11]. The GRIN lens-based common-path (CP) probe only showed a sensitivity up to 44 dB in SDOCT[12]. Although fused ball lenses demonstrated very impressive performance in non-CP Doppler OCT, several main drawbacks of fused lenses for some applications are lower refractive index ($n = 1.48$), teardrop (non-spherical shape), limited diameter (generally less than 500 μm and limits the focal length), and they are fragile, which prevents the imaging lens from being directly exposed to tissue.

In this Letter, we propose a novel sapphire ball lens-based fiber-optic CP-OCT probe applicable for intraocular imaging without dispersion and complex conjugate issue, with sensitivity up to 88 dB. The sapphire ball lens has excellent optical-imaging quality and a

high refractive index of 1.75, which could improve lateral resolution and increase focusing power. The higher index enables the probe to function even when it is submerged in the vitreous gel. This advantage saves the protective sheath or sealed optical window to simplify the probe design and fabrication to minimize its dimension. The relatively perfect spherical shape of the sapphire lenses also reduces the astigmatism and partial coma that generally occurs in fused ball lenses. In addition, a theoretical sensitivity model of CP-OCT was derived and shows that the sensitivity of our probe is up to 88 dB—approaching the theoretical limitation of CP-OCT—which is significantly higher than a recently reported CP-SDOCT probe based on GRIN lens sensitivity of 44 dB [12]. Two 25-gauge prototype common-path fiber probes have been developed; the one with lateral resolution of 11 μm was demonstrated for bovine corneal and retinal imaging *ex vivo* in air and in vitreous gel.

Customized Gaussian beam paraxial ray ABCD matrix simulation [13] (figure not illustrated here) shows that WDs vary with the diameter of the ball lens, wavelengths, and length and type of beam-expanding spacer. Generally, WD at a fixed wavelength is proportional to the diameter of the sapphire ball lens and wavelength. We fabricated two probes in-house to validate the simulation. They were assembled with a single-mode fiber (SMF-28) and a standard 25-gauge hypodermic needle. First, a section of air gap or UV epoxy spacer with refractive index of 1.51 were added between the single-mode fiber distal tip and a sapphire lens with a diameter of 500 μm . Then the air gap or UV epoxy gap were adjusted properly to achieve designed working distance. The reference power is from the fiber distal tip. The WDs were experimentally obtained from the sensitivity falling off of two probes. The parameters of two designs are listed in Table 1.

To the best of our knowledge, no CP-OCT probes have been reported to reach sensitivity of 88 dB [12, 14]. A dual-balanced detector cannot be used for CP-OCT since it will reject the CP-OCT signal as well as other common-mode optical noises. To estimate the optimum performance of CP-OCT with an unbalanced detector, we derived the sensitivity model of CP-OCT by modifying the analysis in prior studies of dual-arm configuration [15–18]. The time-averaged signal power in single port of unbalanced detector of CP-OCT can be expressed as $\langle i_s^2(t) \rangle = (\eta e / hf)^2 P_r P_s$. Here p_r and P_s denote the reference and signal power individually; η is quantum efficiency, e is electron charge, f is optical frequency, h is Planck's constant. The noise power of a single detector contributed by total noises is given as

$$\langle i_n^2(t) \rangle = [4kT/R + 2\eta e^2 / hf * (P_r/2 + P_s/2) + (\eta e / hf)^2 * \text{RIN} * (\zeta * ((P_r/2)^2 + (P_s/2)^2) + P_r P_s / 2)] * \text{BW} \quad (1)$$

where $4kT/R$ represents thermal noise and second term is shot noise. The third terms include RIN (relative intensity noise) noise induced by self-beating and cross-beating noises, ζ is called common-mode rejection ratio, which is 0 dB for common-path OCT and typically –35 dB for balanced detector; BW is the bandwidth. Therefore, the sensitivity of the CP-OCT in dB can be expressed as $\text{Sensitivity} = 10 \log (i_s^2(t) / i_n^2(t))$.

To verify the theoretical sensitivity analysis of CP-OCT, we implemented a CP-SSOCT, which is fully compatible with dual-arm configuration of SSOCT. Probes with similar design can also be used for SDOCT as long as we change the type of single-mode fiber. The reference plane of the probe radiated at the distal end of the single-mode fiber (SMF) was encased within a 25-gauge hypodermic needle and capped with a sapphire ball lens, facilitating its applications in liquid environments. We used a 1310-nm swept source laser with a wide tuning range of 100 nm (Axsun Technologies, Inc.) as the OCT engine, operating at a 50 kHz repetition rate with an axial resolution of 19 μm . Only the positive input port of a dual-balanced detector was used to collect signal in CP-SSOCT. We used the

following experimental parameters for sensitivity measurement: $P_s = 0.7 \mu\text{W}$, $4kT/R = 4.84 \text{ pA}^2/\text{Hz}$, $\zeta = 1(0\text{dB})$, $B = 50\text{MHz}$ and $\text{RIN} = 2.54 \times 10^{-14}/\text{Hz}$ (-135.9dB/Hz) which was the value at 25MHz obtained using a spectrum analyzer. We used an equivalent ND filter of 37.5dB to measure the sensitivity. The experimental sensitivity of the probe in green was compared to theoretical prediction of CP-SSOCT in Fig. 2 in black.

The theoretical sensitivity plot of CP-SSOCT in Fig. 2 in black represents experimentally measured coupling efficiency of 40% which is mainly due to limited effective NA determined by the diameter of ball lens, the distance between fiber distal tip and ball lens center, the immersed media and spherical aberration. Both WD and NA should be increased on the surrounding medium due to increased refractive index of the tissue. The experimental sensitivity in green agrees very well with the theoretical sensitivity of the CP-OCT in black and it is only 8 dB less than the dual-arm bulky imaging head SSOCT when reference arm power is around $20 \mu\text{W}$. The optimal sensitivity is around 88 dB when the reference arm power is between $19 \mu\text{W}$ and $25 \mu\text{W}$. The best sensitivity that CP-SSOCT can achieve is only around $3\sim 4\text{ dB}$ lower than that of traditional dual-arm SSOCT, provided the probe's coupling efficiency increases to 86.5% . The sensitivity of the probe in surrounding tissue depends on the scattering property of the medium, which will deform the wavefront of OCT beam. The probe having an air gap with WD of $415 \mu\text{m}$ is preferable for biological imaging due to better lateral resolution of $11 \mu\text{m}$. Another probe using UV epoxy as spacer is optimal for distance sensing due to its long WD and depth of field.

We performed bovine cornea and retina imaging *ex vivo* both in air and in vitreous gel to assess the probe's performance. The cornea images of Fig. 3A and Fig. 3B were captured by manually scanning the probe tip in air. The sapphire ball lens keeps a distance of $300 \mu\text{m}$ from the corneal apex and around $400 \mu\text{m}$ at the periphery of the cornea in Fig. 3A. Five corneal layers: epithelium, Bowman's layer, stroma, Descemet's membrane, and endothelium, were clearly resolved without any special post-processing. Fig. 3B was obtained at a lateral distance more than 2 mm from the apex. The endothelium layer can still be resolved even at a relatively large incident angle. Fig. 3C was illustrated by manually scanning the probe across the optical nerve head (ONH). Owing to the protected reference plane inside the needle, the left part of Fig. 3C can still be clearly imaged even when the probe is in contact with the tissue. The cross-sectional image far away from the nerve head was also displayed in Fig. 3D, which clearly shows the retinal layer structures. To mimic the clinical environment, the probe was inserted into the vitreous sack of a 34-mm diameter bovine eyeball for imaging and sensing. Considering that the probe was immersed in vitreous gel and that the vitreous gel has high viscosity, the acquired images, as shown in Fig. 3E and F show a notable image quality. These two figures clearly show the retinal layer structures which shows that the probe functions even when submerged in liquid. The dark dots in choroid area of Fig. 3F are hollow blood vessels. Some retina layers were already degenerated since the images were taken approximately 2 hours after the animal was sacrificed at a local farm.

To conclude, we demonstrated a novel sapphire ball-lens probe for endoscopic biological tissue imaging. This is the first demonstration of a sapphire ball lens-based common-path OCT probe capable of imaging in the vitreous gel with sensitivity up to 88 dB , approaching the theoretical limitation of CP-OCT. Its common path performance is significantly better than that of GRIN lens[12] or glass ball lens-based CP-SDOCT due to matched refractive index. This device could be a valuable imaging and sensing tool for ophthalmology, gastrointestinal endoscopy, vascular systems, and brain plaque imaging, and it could potentially substitute a bulk imaging head in some applications. A large diameter sapphire lens around 1 mm can be chosen to improve its working distance at least 5 times with a

decent lateral resolution. The reflection artifacts of the ball lens can be removed with a proper AR-coating.

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Fig. 1. (Color online) Schematic of CP-SSOCT compatible with a dual-arm configuration. A, CP-SSOCT with a lensed probe. The fiber tip functions as a reference plane and there is air gap or UV epoxy between fiber tip and ball lens as spacer. B, The goal of lensed probe is for forward viewing in retina vitrectomy surgery and the probe is attached to the minute surgical forceps.

Fig. 2.

(Color online) Theoretical sensitivity and experimental results. A, Theoretical sensitivity of CP-SSOCT probe is in black with backward optical coupling efficiency of 40%; experimental results are in green; CP-SSOCT with backward coupling efficiency of 86.5% ($1/e^2$ width) is in cyan; traditional dual balanced dual-arm bulky imaging head SSOCT with backward coupling efficiency of 86.5% is in blue.

Fig. 3.

Cow's cornea and retina images (H 2.75 mm by V 2.1 mm) A, Cornea image. We can clearly identify the following five layers: epithelium (1), Bowman's layer (2), stroma (3), Descemet's membrane, (4), and endothelium (5). B, Corneal image far away from the apex. C is the OCT cross-sectional image of optical nerve head. D is away from the optical nerve head and shows clear layer structures of retina. E and F were captured when the probe was immersed in vitreous gel with a length of 32 mm. The probe can still work properly without being damaged when the ball lens contacted the right corner part of retina in Fig. 3E on purpose. All images in Fig. 3 did not have frame average.

Table 1Design parameters of two probes (all units in μm)

Spacer/Length	Theoretical WD	Experimental WD	DOF/Spot size
Air/275	390	415 $\pm$ 5	151/11
Epoxy/169	1197	1221 $\pm$ 15	1478/18