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Sapphire ball lensed fiber probe for common-path optical coherence tomography in ocular imaging and sensing

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Abstract

We describe a novel common-path optical coherence tomography (CP-OCT) fiber probe design using a sapphire ball lens for cross-sectional imaging and sensing in retina vitrectomy surgery. Single mode Gaussian beam (TEM⁰⁰) simulation was used to optimize lateral resolution and working distance (WD) of the common-path probe. A theoretical sensitivity model for CP-OCT was proposed to assess its optimal performance based on an unbalanced photodetector configuration. Two probe designs with working distances (WD) 415 μ m and 1221 μ m and lateral resolution 11 μ m and 18 μ m, respectively were implemented with sensitivity up to 88dB. The designs are also fully compatible with conventional Michelson interferometer based OCT configurations. The reference plane of the probe, located at the distal beam exit interface of the single mode fiber (SMF), was encased within a 25-gauge hypodermic needle by the sapphire ball lens facilitates its applications in bloody and harsh environments. The performances of the fiber probe with 11 μ m of lateral resolution and 19 μ m of axial resolution were demonstrated by cross-sectional imaging of a cow cornea and retina *in vitro* with a 1310nm swept source OCT system. This probe was also attached to a piezoelectric motor for active compensation of physiological tremor for handheld retinal surgical tools.

Keywords

optical coherence tomography; retina imaging; fiber probe

I. INTRODUCTION

Optical coherence tomography is becoming widely used tools for non-destructive cross-sectional imaging within biological tissues[1–7]. A single mode fiber can be lensed with state-of-the-art micro-optics to form a light beam with spot size around 20 μ m~50 μ m in gastrointestinal endoscopy, coronary artery imaging and needle based Doppler OCT[8]. The commonly used lensing components in fiber-optic microprobes are gradient-index lens[3, 9–12], drum lens[13], fiber fused ball lens[8], and special liquid forming ball lens. However, there are three challenges remained using micro-optical imaging for clinical surgery in harsh environment. Firstly, almost all the current fiber-optic OCT probes need a separate reference arm, which is very inconvenient for changing the imaging probe during some sophisticated surgery due to the mismatch of optical path length, dispersion compensation and complex conjugate issue. Secondly, the fiber-optic probes are expected to keep in tact even if they contact the rigid specimens. For example, the fiber lens can scratch rigid sample surface such as cochlear temporal bone without being damaged during robotic assistant surgery.

Thirdly, a common path probe should have an isolated reference plane to prevent blood block the reference power in the case of retinal surgery or cardiovascular coronary imaging. An optimal probe should have a protected reference plane and smooth curved rigid imaging surface with longer working distance so as to prevent scratching fine structure of tissues such as retina[14].

In this paper, we proposed a sapphire ball lens based fiber-optic common path OCT (CP-OCT) probe without complex conjugate issue with sensitivity up to 88dB, meeting all three challenges and works for both spectral domain OCT (SDOCT) and swept source OCT (SSOCT). Sapphire ball lens has excellent optical imaging quality, very robust, high refractive index ($n=1.75$), and smooth surface. Two prototypes of 25-gauge common-path fiber probe have been developed and the one with lateral resolution of $11\mu\text{m}$ was demonstrated for cow corneal and retinal imaging *in vitro*. Gaussian beam paraxial ray ABCD matrix[12] was employed to simulate beam propagation inside probe and provide the proper design parameters.

II. METHODS

Sapphire ball lens has excellent optical imaging quality and owns extraordinary stiff smooth surface. Gaussian beam paraxial ray ABCD matrix[12] was employed to simulate beam propagation inside probe and provide the proper design parameters. The beam propagation of Gaussian beam inside probe can be approximated by a paraxial optical system. Generally three main parameters, complex parameter $y(z)$, the radius of curvature $R(z)$ and beam waist are used to describe a Gaussian beam by $1/y1 = 1/R1(z) - i \lambda / (\pi \omega^2)$. Here λ is the free space central wavelength and n is the refractive index. The y parameters, $y1$ and $y2$, of the Gaussian beam at the exit plane of the single mode fiber (SMF, TEM₀₀ mode) fiber and output plane after sapphire ball lens can be related by a paraxial optical (A, B; C, D) matrix of the probe $y2 = (Ay1 + B)/(Cy1 - D)$. The mode field radius ($4.6\mu\text{m}$) of the SMF is used as the input Gaussian waist instead of the fiber core. Using the Gaussian beam conditions $R2(z) \rightarrow \infty$ at focal plane, the relationships relating focal length, also known as working distance from the outer surface of the ball lens, beam spot size, wavelength, beam expanding gap between SMF and ball lens could be numerically obtained. The simulated working distances vary with the diameter of ball lens, wavelengths and length and type of beam spacer. Generally WD at fixed wavelength is proportional to the diameter of sapphire ball lens and wavelength, which is clearly demonstrated from variances of the WD with diameters of $500\mu\text{m}$, $800\mu\text{m}$ and $1000\mu\text{m}$, respectively. For example, WD can be up to 4.5mm in the case of $D=1000\mu\text{m}$ at wavelength of $1310\mu\text{m}$, which is enough for most of the endoscopic imaging and cardiovascular intracoronary imaging. The numerical simulations also study the impact of chromatic aberration on focal length. The differences of WD caused by chromatic aberration compared to 1310nm are $44.9\mu\text{m}$ for 1365nm and $-44.8\mu\text{m}$ at 1255nm respectively. The difference is only about 3.7% of WD with $1197\mu\text{m}$ and does not have big impact on depth of field (DOF). One prototypes of 25-gauge fiber probe has been developed with lateral resolution of $11\mu\text{m}$. The system configuration is illustrated in Fig. 1.

Two probes were fabricated in-house to validate the simulation. They were assembled with a single mode fiber (SMF-28) and a standard 25-gauge interventional needle. A section of air and UV epoxy spacer with refractive index of 1.51 were added between the single mode fiber end and a sapphire ball lens with a diameter of $500\mu\text{m}$, which is larger than that of usual non-core fiber fused ball lens. The parameters of two designs are listed in Table 1. The spacers for two probes have similar optical path length of $275\mu\text{m}$ with air and of $255\mu\text{m}$ with UV curing epoxy. However, the difference of their WD is $807\mu\text{m}$. It means that WD is not only determined by spacer gap and also related to the matching condition of refractive index at transition interfaces of the probe.

Another main concern is the sensitivity of the CP-OCT probe. To the best of our knowledge, there were no CP-OCT probes reported to reach more than 80dB sensitivity. Dual balanced detector cannot be used for CP-OCT since it will reject CP-OCT signal as well as other common mode optical noises. To estimate the optimum performance of CP-OCT with unbalanced detector, we derive the signal to noise (SNR) ratio of CP-OCT by modifying the analysis in prior works [15–17]. The time averaged signal power in single port of unbalanced detector of CP-OCT can be expressed as $\langle i_s^2(t) \rangle = (\eta e / \eta f)^2 P_r P_s$. Here P_r and P_s denote the reference and signal power individually, η is quantum efficiency, e is electron charge, h is Plank's constant. The noise power of a single detector contributed by total noises is given as below,

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

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 0dB for common-path OCT and typically -35dB for balanced detector; BW is the bandwidth. Therefore, the sensitivity of the CP-OCT in dB can be expressed as $\text{Sensitivity} = 10 \log (i_s^2(t) / i_n^2(t))$.

III. RESULTS AND DISCUSSION

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 50kHz repetition rate with an axial resolution of 19 μ m. 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: $P_s = 0.7\mu$ W, $4kT/R = 4.84pA^2/Hz$, $\zeta = 1$ (0dB), $B = 50MHz$ and $RIN = 2.54 * 10^{-14}/Hz$ (-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 experimental data agrees very well with the theoretical sensitivity of the CP-OCT and it is only 8dB less than traditional SSOCT when reference arm power is around 20 μ W. The optimal sensitivity is around 88dB when the reference arm power is between 19 μ W and 25 μ W [18]. The best sensitivity that CP-OCT can achieve is only 3dB lower than that of traditional SSOCT provided its coupling efficiency increasing to 86.5%.

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 μ m from the corneal apex and around 400 μ 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.

The probe was intelligently integrated with a mini-servo PZT-motor as a smart surgical tool to reduce tremor and motion tracking. To show that the new probe can be highly effectively in the smart surgical tool setting, we used a phantom to mimic the inner limiting membrane, which has a high reflectivity to assess its performance. Experimental result shows that the lensed probe could sense the phantom surface over a wide viewing angle and can be used to maintain a stable distance of tool top to target surface up to 3.5mm using a PZT motor and active proportional-integral-derivative (PID) control algorithm. The amplitude of the tremor was reduced from 71.7 μ m to 12.7 μ m. In additional, the working distance of lensed fiber is much longer than that of bare fiber sensor, which is only stable around 1.0 mm. The results are illustrated in Fig. 4.

IV. CONCLUSION

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 performance is significantly better than that of GRIN lens[19] or glass ball lens-based CP-SDOCT. 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 and 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-swept source optical coherence tomography (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 an air gap or UV epoxy between the fiber tip and ball lens as spacer. B. Goal of lensed probe is for forward viewing in retina vitrectomy surgery, and the probe is attached to the minute surgical forceps.

Fig. 2.

Theoretical sensitivity and experimental results. A, Sensitivity of CP-OCT is in black with backward optical coupling efficiency of 40%; experimental results are in green; CP-OCT with backward coupling efficiency of 86.5% ($1/e^2$ width) is in cyan; traditional balanced SSOCT with backward coupling efficiency of 86.5% is in blue. B, SNR and falling off of two probes. The probe with a focal length of $415\mu\text{m}$ is shown in green and $1221\mu\text{m}$ in black. The SNR and plots were directly obtained from purely single A-scan without any A-scan average.

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. All images in Fig. 3 did not have frame average.

Fig. 4.

A: Distance between the lensed fiber probe tip and the phantom surface. At 2.0mm, the compensation PZT motor is off and OCT signal shows a strong oscillation introduced on purpose. The other segments are with compensation on so that the probe follows the movement.

Table 1 Design 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
UV/169	1197	1221 $\pm$ 15	1478/18