Full-range Fourier-domain optical coherence tomography imaging probe with a magnetic-driven resonant fiber cantilever

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Full-range Fourier domain optical coherence tomography imaging probe with a magnetic-driven resonant fiber cantilever

Kang Zhang
Yong Huang
Jin U. Kang
Full-range Fourier domain optical coherence tomography imaging probe with a magnetic-driven resonant fiber cantilever

Kang Zhang
General Electric Global Research Center
1 Research Circle - K1 3A-42
Niskayuna, New York 12309
E-mail: kangzhang2011@gmail.com

Yong Huang
Jin U. Kang
Johns Hopkins University
Department of Electrical and Computer Engineering
3400 North Charles Street, Barton Hall
Baltimore, Maryland 21218

Abstract. In this work, we developed a full-range Fourier domain optical coherence tomography (FD-OCT) imaging probe with a magnetic-driven resonant fiber cantilever. A galvanometer-driven reference mirror provides a linear phase modulation to each M-scan/B-scan frame to enable complex conjugate-free, full-range FD-OCT imaging. A fiber cantilever inside the probe is driven by a low-voltage miniature magnetic transducer. The fiber cantilever scans at its resonant frequency synchronized to the phase modulation induced by the reference mirror. Using a CCD line-scan camera-based spectrometer, we demonstrated real-time full-range FD-OCT at 34 frame/s (1024 pixel lateral × 2048 pixel axial). © 2011 Society of Photo-Optical Instrumentation Engineers (SPIE). [DOI: 10.1117/1.3645088]

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1 Introduction

Optical coherence tomography (OCT) is a noninvasive imaging modality capable of providing depth-resolved images of biological tissues with micrometer level resolution.1 These features make OCT systems highly suitable for applications in microsurgical guidance and intervention.2–10 In the last few years, ultrahigh speed Fourier domain optical coherence tomography (FD-OCT) has generally achieved >100,000 line/s A-scan imaging rate,11–15 and even megahertz OCT is available in acquisition,16,17 reconstruction and visualization,18 which is very necessary for interventional applications.

Compared to other imaging modalities such as CT, ultrasound, and MRI, which have already been widely used in image-guided intervention, OCT has the disadvantage of shallower imaging penetration depth. As a solution, endoscopic OCT has been proposed for intrabody imaging.19 A wide range of miniature endoscopic OCT probes have been developed to provide high resolution imaging while being flexible and integratable with medical devices such as a rotary OCT imaging needle20 and balloon catheter,21 paired-angle-rotation scanning probe,22 polymer-based scanning cantilever,23 piezoelectric scanning mirror,24 MEMS scanning mirror,25,26 piezoelectric,27 and magnetic force-driven resonant fiber scanner.28,29 With the rapid development in OCT technologies, in recent years endoscopic OCT also underwent a transition from traditional low-speed time-domain mode to high-speed Fourier domain mode.30,31

However, as in the regular bulky-scanner–based systems, the FD-OCT imaging probes can also suffer from a complex-conjugate-free FD-OCT with a forward-viewing miniature resonant fiber-scanning probe. A galvanometer-driven reference mirror provides a linear phase modulation to the A-scans within one frame and the simultaneous B-M-mode scanning are implemented. Inside the probe, a fiber cantilever is driven by a low-voltage magnetic transducer synchronized to the reference mirror scanning. Using a CCD line-scan camera-based spectrometer, we demonstrated real-time full-range FD-OCT imaging with doubled imaging range at 34 frame/s.

2 Probe Development

First we designed a forward-viewing miniature resonant fiber-scanning probe based on a low-voltage miniature magnetic transducer, as shown in Fig. 1(a). Inside the probe, a miniature magnetic transducer with 5-mm diameter and 2-mm thickness is used to drive a single-mode fiber cantilever, which is attached to the diaphragm of the transducer. The fiber cantilever scans within the plane perpendicular to the diaphragm at its resonant frequency. Two scanning lenses are placed after the fiber to image the beam across the sample. The fiber tip is placed at the focal plane of scanning lens 1 so that the distal imaging plane can be located at the focal plane of scanning lens 2. This lens setup minimizes the imaging field distortion and avoids the need for the geometrical correction during the image processing stage. Figure 1(b) shows the ZEMAX simulation of the scanning lens set, where we combined two achromatic lenses to form scanning lens 1 (proximal end), and used another one as scanning lens 2 (distal end). All three lenses are type AC050-008-B, 5-mm diameter, f = 7.5 mm, Thorlabs, therefore a telemetric system with an approximate magnification of 1:2 is formed by the three lenses. As shown in the optimized ZEMAX simulation in Fig. 1(b), with the fiber scanning range of 1 mm, an imaging range of 1.94 mm is obtained at the scanning plane of the probe. The effective imaging numerical aperture (NA) is simulated to be about 0.07. The spot size variation is simulated by ZEMAX as in Fig. 1(c), which indicates a resolution...
below 30 μm over the scan range without substantial focusing distortion. In the actual system, the scanning fiber tip is angle-cleaved by ~8 deg to minimize Fresnel reflection and the scanning range is set to 1 mm by adjusting the amplitude of the function generator input to the transducer, as shown in Fig. 1(d). A prototype of the probe is shown in Fig. 1(e), which can be further miniaturized by shortening the scanning fiber, using a smaller transducer, and using smaller scanning lenses.

3 System Configuration

Then, the probe is integrated into a spectrometer-based FD-OCT system, as shown in Fig. 2. A 12-bit CCD line-scan camera (EM4, e2v, USA) works as the linear detector of the OCT spectrometer. A superluminescence diode (λ₀ = 825 nm, Δλ = 70 nm, Superlum, Ireland) was used as the light source, which provided a measured axial resolution of approximately 5.5 μm in air using the common-path interferometer setup. The minimum line period is camera-limited to 14.2 μs, which corresponds to a maximum line rate of 70,000 line/s. A quad-core Dell T7500 workstation was used to host a frame grabber (PCIe-1429, National Instruments) and to implement the data processing. A galvanometer-driven mirror is placed at the end of the reference arm. The fiber-scanning probe and the reference mirror are driven by a dual channel function generator and synchronized with the frame grabber.

To induce a phase modulation, as in Fig. 2, a sinusoidal wave (CH1) was sent to drive the magnetic transducer and a symmetrical triangle wave (CH2) for the galvanometer mirror. The scanning position of the probe was experimentally adjusted so that it is in phase with the reference mirror position and the frame grabber triggering signal, as shown in Fig. 3. The rising and the falling slopes of the reference mirror’s triangle motion applies linear phase modulation to the odd and even image frames, respectively. To fully utilize the 70 kHz line speed of the CCD camera, the resonant frequency of the cantilever was set to ~34 Hz by experimentally adjusting the fiber length. By setting 1024 A-scans to form each frame and scanning frequency to 34 Hz, the imaging speed was 34.8 kHz for single-way scanning and 69.6 kHz for dual-way scanning. The final fiber length was around 55 mm, which can be made shorter by putting a weight on the cantilever tip.

Figure 4 illustrates the galvanometer-driven reference mirror induced phase modulation; using the small-angle approximation, the displacement of the reference position is d = RΔα, where R is the distance from the beam spot to the pivoting point O, Δα is the scanning angle. Then the inter-A-scan phase shift is given by,

$$\Phi = 2\pi * 1 \frac{d}{\lambda_0} \frac{1}{N} = \frac{4\pi R \Delta\alpha}{N\lambda_0},$$

and λ₀ is the central wavelength of the light source. N is the number of A-scans within each frame. In our setup, with λ₀ = 825 nm, we set R = 3 mm, Δα = 2°, and N = 1024.

Figure 5(a) shows an M-scan frame of 1024 A-scans obtained by keeping the probe focused on a fixed mirror, from which we deduced the actual reference displacement value of d = 109 μm. This corresponds to an experimental

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Fig. 1 (a) Probe design. (b) ZEMAX simulation of lens set. (c) ZEMAX simulation of spot size. (d) Fiber scanning of 1 mm. (e) A probe prototype.

Fig. 2 System configuration: CCD, CCD line-scan camera; G, grating; L1, L2, L3, achromatic lenses; CL, camera link cable; COMP, host computer; C, 50:50 broadband fiber coupler; PC, polarization controller; GV, galvanometer with reference mirror; MT, magnetic transducer; SMF, single-mode fiber; SL1, SL2, scanning lens; SP, sample; FG, function generator.

Fig. 3 Positions of probe and reference during the phase modulation.

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phase shift of $\Phi = 1.62$, which is very close to the optimal phase shift of $\pi/2$. With a minimum line period of 14.2 $\mu$s, a maximum carrier frequency of 11.4 kHz was obtained, which corresponds to an axial motion speed of 7.3 mm/s. Figure 5(b) presents the modulation in the amplitude of the reference within each image frame. Figure 5(c) shows normalized reference spectrums corresponding to three different A-scans within each frame. As one can see, the amplitude of the reference at the edges of the image frame is about 60% of that in the middle, while the spectral shape remains relatively unchanged and the intensity modulation is compensated accordingly during the imaging processing. The phase modulation can also be implemented by a piezoelectric stage-driven mirror, a fiber stretcher, or an electro-optic phase modulator. In the present stage, we focus on the two-dimensional (2D) cross-sectional imaging via one-dimensional (1D) scanning. As a further step for three-dimensional volumetric imaging, the 2D scanning by a single magnetic actuator could be realized through the asymmetric cantilever approach.

4 Image Processing

As shown in Fig. 3, within a single odd/even frame, a linear phase modulation $\varphi(t|x) = \beta \delta(t|x)$ was applied to each M-scan/B-scan’s 2D interferogram frame $I(t|x, \lambda)$. Here, $t|x$ indicates time/position index of A-scans in each M-scan/B-scan frame. By applying Fourier transform along the $t|x$ direction, the following equation can be obtained:

$$F_{t|x} \rightarrow f|u[I(t|x, \lambda)] = |E_s|^2 \delta(f|u) + \Gamma_{f|u}[F_{t|x} \rightarrow f|u[E_r(t|x, \lambda)]]$$

$$= F_{t|x} \rightarrow f|u \left[ E_s^* \delta(f|u + \beta) + F_{t|x} \rightarrow f|u \left[ E_s(t|x, \lambda) E_r^*(\lambda) \right] \otimes \delta(f|u - \beta) \right],$$

(2)

where $E_s(t|x, \lambda)$ and $E_r(\lambda)$ are the electrical fields from the sample and reference arms, respectively. $\Gamma_{f|u}[\cdot]$ is the correlation operator. The first three terms on the right-hand side of Eq. (2) present the direct current (dc) noise, autocorrelation noise, and complex-conjugate noise, respectively. The last term can be filtered out by a proper bandpass filter in the $f|u$ domain and then convert back to the $t|x$ domain by applying an inverse Fourier transform along the $f|u$ direction. The complete data processing can be divided into following steps:

1. The dc removal and intensity compensation according to Fig. 5(b);
2. Remapping the spectrum from $\lambda$ domain to $k$ domain;
3. Fourier transform along the $t|x$ domain to the $f|u$ domain;
4. Bandpass filter in the $f|u$ domain by a super-Gaussian filter;
5. Inverse Fourier transform along the $f|u$ domain back to the $t|x$ domain;
6. Numerical dispersion compensation by phase correction;
7. Fourier transform along the $k$ domain to obtain depth information;
8. Correct the image distortion caused by the phase modulation;
9. Correct the image distortion caused by the sinusoidal scanning.
In particular, here step 6 is realized by adding a phase correction term \( \Phi = -a_2 (\omega - \omega_0)^2 - a_3 (\omega - \omega_0)^3 \) to the complex spectrum obtained after steps 3 to 5, which serve as a modified Hilbert transform, where \( a_2 \) and \( a_3 \) are preoptimized values according to the system properties.\(^{37}\)

We then processed the M-scan data frame in Fig. 5(a) following steps 1 to 8, and the results of the last two steps are shown in Fig. 6. Here, step 4 is not presented since no B-scan was conducted. As clearly shown in Fig. 6(b), the complex-conjugate artifact was removed and the phase distortion by mirror displacement was also corrected, presenting a flat line in only one side of the zero-delay line.

A single A-scan is extracted from the frame in Fig. 6(a), as indicated by the red (vertical) line, and shown in Fig. 7(a). The complex-conjugate artifact suppression was better than 55 dB. Figure 7(b) compares the dispersion compensated and uncompensated A-scan profiles. With the numerical
dispersion compensation, we have obtained a resolution (5.6 μm) very close to the premeasured value (5.5 μm).

5 Imaging Tests

In the imaging test, the probe was operated under the single-direction scanning mode at 34 frames/s, and only the odd frames were acquired and processed with an image size of 1024 pixel lateral by 2048 pixel axial. Figures 8(a) and 8(b) show the image results of an IR card without and with the reference-arm phase modulation as a comparison. Under the phase modulation mode, the complex-conjugate artifact was effectively suppressed and the autocorrelation lines due to internal multiple reflection [the horizontal lines in Fig. 8(a)] were also eliminated.

Finally, we conducted in vivo human finger imaging using the scanning probe with the same scanning protocol as depicted in Fig. 8(b). Figure 9(a) presents the coronal scans of the finger tip, where the epithelial structures such as stratum corneum, stratum spinosum, and sweat duct are clearly visible. Figure 9(b) shows the sagittal scan of finger nail fold region, showing the major dermatomic structures such as epidermis, dermis, nail fold, nail root, and nail bed.

6 Conclusion

In this work, a full-range FD-OCT imaging probe with a magnetic-driven resonant fiber cantilever was developed. The simultaneous B-M-mode phase modulation was implemented by scanning the reference mirror synchronized to the sample arm. The complete image processing was presented and in vivo human finger images were obtained with the complex-conjugate artifact removal. This method can also be generally applied to other types of FD-OCT imaging probes to eliminate the complex-conjugate artifact and double the imaging range.

Acknowledgments

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References


Kang Zhang received a BS degree in physics from Nankai University, Tianjin, China, in 2007, and MS and PhD degrees in electrical and computer engineering from Johns Hopkins University, Baltimore, Maryland, in 2009 and 2011, respectively. He is currently a researcher in the GE Global Research Center, New York. His research interests include optical sensing and imaging, interventional imaging, smart surgical devices, and GPGPU applications.

Yong Huang received a BS degree in physics from Peking University, Beijing, China, in 2009. He is currently a PhD candidate in the Department of Electrical and Computer Engineering, Johns Hopkins University, Baltimore, Maryland.

Jin U. Kang received a PhD degree in electrical engineering and optical sciences from the University of Central Florida, Orlando, in 1996. From 1996 to 1998, he was a research engineer with the United States Naval Research Laboratory, Washington, DC. He is currently a professor and chair in the Department of Electrical and Computer Engineering, Johns Hopkins University, Baltimore, Maryland. His research interests include fiber optic sensors and imaging systems, novel fiber laser systems, and biophotonics.