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Endoscopic Optical Coherence Elastography Using Acoustic Radiation Force and Bending Vibration of Optical Fiber

音響放射力と光ファイバのたわみ振動を用いた光コヒーレンスエラストグラフィ内視鏡

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1. Introduction

Optical coherence tomography (OCT) is an optical imaging technique that provides high-resolution cross-sectional image for clinical and biological field, and the OCT systems have a depth resolution of less than 10 μm. Recently, measurement of hardness using endoscopic OCT is required [1].

We have proposed the high-speed optical scanner in simple structure using bending vibration of an optical fiber for scanning of the measurement light from OCT system to sample tissue[2,3]. In this report, we investigate the miniaturization of the endoscope, and demonstrate endoscopic OCT elastography by combining it with acoustic radiation force .

2. Configuration of endoscopic elastography

The proposed system for endoscopic OCT elastography is illustrated in **Fig. 1**. Deformation of tissue in water is caused with the use of acoustic radiation force which is induced by the difference of acoustic energy density at the interface of the propagating media using a perforated focused transducer. The strain is slowly relaxed after removal of the force, and its behavior is measured by OCT. The displacement and the strain are calculated and imaged with the cross-correlation detection using the OCT images before and after applying the force. The distance is required to be shorter to suppress the light attenuation in water. An optical scanner probe is inserted into the hole of the transducer, and the output light from the probe is forward scanned. Two-dimensional image is obtained by combining the depth scan based on the OCT and the lateral scan of the scanner probe.

3. Endoscopic OCT probe

As shown in Fig. 1, a single-mode optical fiber with the length of resonance condition is attached to the center of a cylindrical piezoelectric transducer [2,4]. The cylindrical transducer has four outer electrodes, and the voltages with a phase shift are applied between the adjacent electrodes to excite a circular [2] or linear [3] vibration at the fiber end. The optical fiber is vibrated in the bending cantilever mode at the resonant frequency, and the output light from the fiber end is collimated by a lens. The collimated light is forward scanned in lateral direction of the tissue sample.

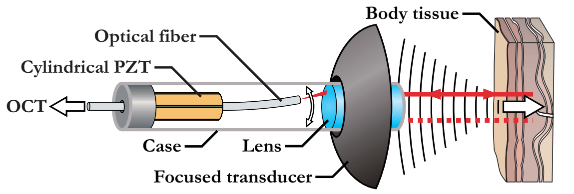
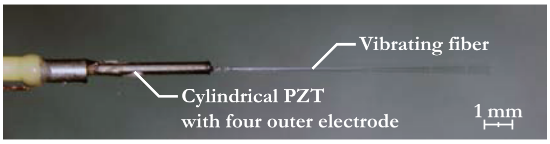


Fig. 1 Basic configuration of the endoscopic OCT-based elastography using the focusing transducer and bending vibration of the optical fiber.



(a) Bending vibration of the optical fiber.

(b) Overview of the endoscopic optical scanner.

Fig. 2 A prototype of the endoscopic OCT probe.

A prototype of the scanner probe is shown in **Fig. 2**(a). The optical fiber of 0.125 mm in diameter without coating is attached to the center of the cylindrical PZT element (outer diameter 0.4 mm, inner diameter 0.2 mm and length 5 mm). In this prototype, the resonant length of the optical fiber in the first bending mode of the cantilever beam is designed to be 10 mm at the frequency of 1 kHz. The end of the fiber is cleaved at 8° angle to suppress the effect of Fresnel reflection at the boundary between the fiber and the air. The collimating lens is a graded index (GRIN) lens of 1 mm in diameter with anti-reflection coating. The PZT element and the

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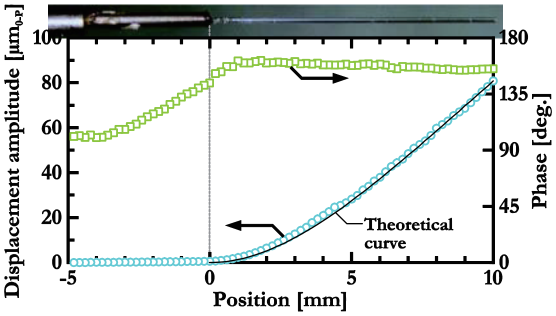


Fig. 3 Displacement distribution along the optical fiber in the first bending mode at 985 Hz.

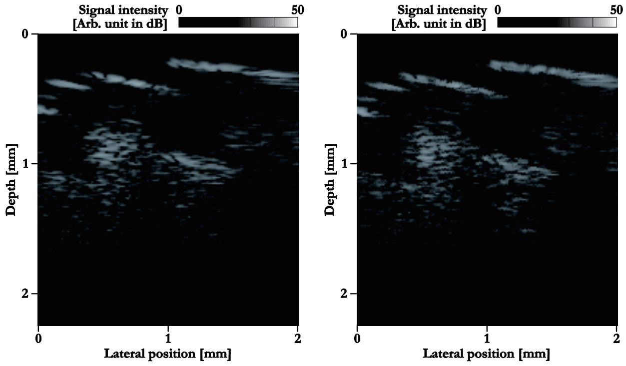
fiber are encased in a stainless-steel tube of 1 mm in diameter, and the GRIN lens is attached to the end of the tubular case as shown in Fig. 2(b). The length of the OCT probe is 19.5 mm.

The vibration displacement distribution along the fiber without the tubular case was measured using a laser Doppler velocimeter at 985 Hz as shown in **Fig. 3**. The measurement was conducted for a sinusoidal voltage of 10 $V_{P\text{-}P}$ with the phase shift of π across one pair of electrodes of the four outer electrodes of the cylindrical PZT. The displacement distribution agrees well with the beam theory for a cantilever. The scan width in lateral direction of the incident light is 2 mm on the tissue 5 mm away from the end surface of the lens.

4. Strain imaging

A focused transducer which has outer diameter of 10 mm, inner diameter of 3 mm and radius of curvature of 15 mm is set in the degassed water. The transducer is driven by sinusoidal burst waves of 10,000 cycles at 2.2 MHz. Finger pad as a sample tissue is set at the focal position of the transducer. The OCT probe is located in the hole of the transducer, and the distance between the end of the probe and the sample is 5 mm.

In this experiment, the developed optical scanner probe are installed in a commercial OCT system (IV-2000, Santec Co.) that consists of a wavelength swept light source and a fiber-based Mach-Zehnder interferometer. The wavelength scanning rate is 20 kHz at the center wavelength of 1330 nm, and the spectral bandwidth of 110 nm results in the depth resolution of 9 μm in water. A sample object is constantly scanned by the wavelength sweep of the light source in the depth direction during the lateral scanning by the vibrating fiber. A cycle of the vibration consists of up- and down-scan, and the lateral scan rate is 1970 scans/sec. Almost 10 depth-scans are obtained for one lateral scan. As the depth scanning frequency is not equal to the integral multiple of the lateral scanning frequency, the number of scanned lines is increased in proportion to the number of lateral scans. In this experiment, a two-dimensional tomographic image was plotted using scanning lines of 400 during 40 cycles, and the frame rate was 40 fps.



(a) Stationary condition. (b) During deformation.

Fig. 4 Two-dimensional OCT images of finger pad.

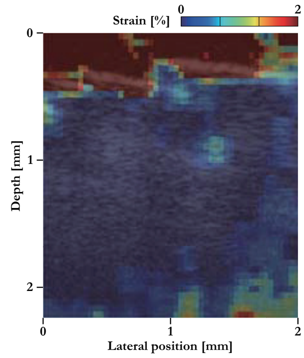


Fig. 5 Longitudinal strain distribution measured by the endoscopic OCT elastography.

The OCT images of the finger pad at the stationary condition and during deformation are shown in **Fig. 4**. Then, two-dimensional displacement vectors are calculated by the cross-correlation between the two images. The maximum displacement was found to be 50 μm in the depth direction. The strains are calculated using the adjacent displacement vectors. The strain image is obtained as shown in **Fig. 5**.

5. Conclusions

The endoscopic OCT elastography using the combination of vibrating fiber and acoustic radiation force was investigated. The strain imaging of finger pad was demonstrated.

Acknowledgment
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