Low-voltage polymer-based scanning cantilever for *in vivo* optical coherence tomography

Yuli Wang, Mark Bachman, and Guann-Pyng Li

Department of Electrical Engineering and Computer Science, University of California, Irvine, Irvine, California 92697

Shuguang Guo, Brian J. F. Wong, and Zhongping Chen

Beckman Laser Institute, Department of Biomedical Engineering, University of California, Irvine, Irvine, California 92612

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Novel hand-held optical coherence tomography (OCT) probes with polymer cantilevers have been developed for clinical oral and skin imaging. An electroactive ionic polymer–metal composite cantilever was used to generate 3-mm transverse scanning movement of an optical fiber with applied 2-V linear alternating voltage at 1 Hz. Low driving voltage ensures safety. Two different optical designs achieve both forward and sidewise scanning and make it possible to image everywhere within the human oral cavity. *In vivo* OCT imaging of the human tongue is demonstrated. © 2005 Optical Society of America

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Optical coherence tomography (OCT) is a noninvasive imaging technique that offers high-resolution cross-sectional imaging of highly scattering media such as biological tissue.[1] OCT has been applied to image subsurface structures in skin, retina, **blood** vessels, oral cavities, gastrointestinal tracts, etc. A key component of an OCT system is the actuator, which facilitates linear scanning by an optical beam across a target to create a two-dimensional image. Several conventional electrical actuation methods were developed previously, including the use of a piezoelectric cantilever to move an optical fiber,[2,3] the use of a thermoelectric actuator[4] and an electrostatic actuator[5] to swing a microelectromechanical mirror, and the rotation of electromagnetic devices (e.g., a galvanometer shaft[6] and a microelectromechanical motor[7]). Although these actuation methods have met the linear scanning requirement, they typically require a relatively high driving voltage[2–5] or a complicated instrumental structure.[6,7] For clinical implementation of OCT systems, such as endoscopic and intravascular applications, low voltage is preferred for safety. In addition, the actuator must be simple in structure, rugged in handling, and small in size to work well within an endoscope. In this Letter we report a novel, compact polymer-actuator-based OCT scanning probe. The polymer actuator has a low driving voltage (2 V) and meets the requirement for incorporation into an endoscope.

Electroactive polymers (EAPs) exhibit large displacement in response to external electrical stimulation. EAPs behave similarly to biological muscles in their low density, short response time, resilience, and large actuation strains. As a result EAPs have acquired the nickname "artificial muscles".[8] Among various classes of EAP, ionic polymer–metal composites (IPMCs) are especially interesting because they show large deformation in the presence of very low voltage (1–2 V).[9–11] An IPMC typically consists of a thin perfluorinated polyelectrolyte sandwiched by electroplated platinum or gold on both sides [Fig. 1(a)]. When external voltage is applied across the IPMC, mobile positive ions are drawn to the cathode side by an electric field. As a result the cathode side expands with respect to the anode, causing an overall bending deformation of the IPMC until it reaches saturation [Fig. 1(a)]. When alternating voltage is applied, the IPMC undergoes a bending vibration at the same frequency as the applied voltage. Although a humid environment is required for the best operation, the IPMCs can operate in a dry environment after proper encapsulation. Figure 1(b) shows the maximum bending of an IPMC strip (Environmental Robots, Inc., Albuquerque, N.Mex.) on application of ± 2 V dc in air. Compared with conventional electrical actuators,

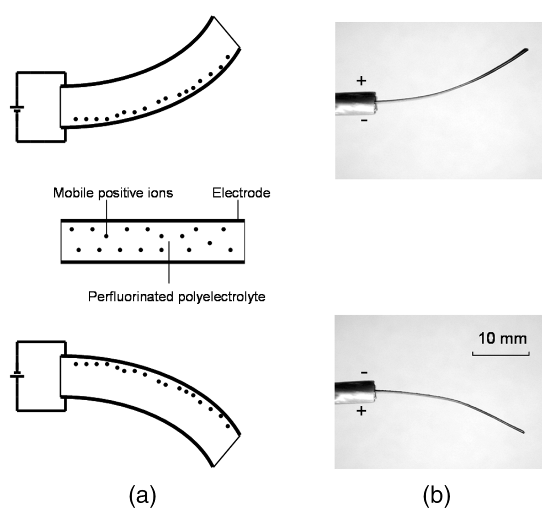

Fig. 1. (a) Schematic of the IPMC actuator and its mode of operation. (b) Photographs of an IPMC strip that shows large deformation (as much as 15 mm) in the presence of a voltage of ± 2 V DC. (The 0.3-mm-thick IPMC was purchased from Environmental Robots, Inc., and cut into a 2 mm × 30 mm strip.)

IPMCs are unique in their low operating voltage and simple structure (i.e., no conventional mechanical moving parts).

An optical fiber is directly attached to the IPMC actuator such that the bending vibration of the IPMC can generate a scanning movement of the optical beam. Thus hand-held OCT probes were incorporated with IPMC cantilevers, and their schematic diagrams are shown in Fig. 2. In one design [Fig. 2(a)] the 0.3-mm-thick IPMC plate was cut into a 30 mm × 2 mm strip by a sharp scalpel. Then a polytetrafluoroethylene (PTFE) hollow tube (Small Parts, Inc.; inside diameter, 0.046 cm) was fixed on the tip of the IPMC strip by epoxy glue. This PTFE tube was used to house an optical fiber with a 0.7-mm-diameter gradient-index lens glued to the fiber tip. Using a gradient-index lens attached to the scanning fiber is better than scanning the fiber before a fixed lens because the latter procedure will result in coma aberration. The IPMC cantilever and the optical fiber were inserted into a one-end-open rectangular glass tube with cross-sectional dimensions of 5 mm × 10 mm. The use of a rectangular glass tube instead of a round tube ensures that the imaging beam will pass through the tube with minimal signal loss and distortion. Two electrodes of the IPMC were then attached to electrical wires. Finally, medical-grade epoxy sealant was used to seal the open end of the tube to protect the IPMC cantilever from the environment. Actuation bending of the IPMC cantilever causes displacement of the distal fiber tip in the transverse direction, which results in a transverse displacement of the beam focus in the objective plane in the front of the probe. In another design [Fig. 2(b)] an additional 0.7-mm prism was glued onto the gradient-index lens to reflect the optical beam, resulting in beam scanning in the lateral side of the probe. The first design, which is the forward scanning probe, is suitable for imaging the *in vitro* samples and *in vivo* external parts of the human body (e.g., skin), whereas the second design, which is the sidewise scanning probe, is ideally suitable for imaging the part of human oral cavity that is difficult to approach with the first probe. In both designs the focus of the imaging beam is located ~1 mm away from the outer wall of probes. The probes are simply placed in contact with samples to be visualized during OCT imaging.

In the OCT system, a linear scanning movement of the fiber beam is required for undistorted OCT images. To measure the bending movement of the optical beam controlled by the IPMC on alternating voltage (2 V and 1 Hz) we used a Kodak megaplus CCD camera to record the entire process of beam movement at 60 frames/s. A ratio of beam displacement to time was then obtained through video analysis. We found that the scanning movement is nonlinear when regular voltage waveforms (e.g., square, triangular, sinusoid, saw-toothed) are applied. As a result, the OCT images obtained on these waveforms are distorted. When the modified voltage waveform shown in Fig. 3 is applied, the movement is linear, with an average linearity (R^2) of 0.998. The OCT images obtained on this waveform is undistorted. The linearity of the IPMC bending under the modified voltage waveform can be explained by **its** actuation mechanism, as shown in Fig. 1(a). Bending of the IPMC is caused by a voltage-induced shift of mobile positive ions contained in its body: the larger the applied voltage, the more the positive ions are drawn to the cathode side and the larger the deflection of the IPMC strip. The IPMC strip bends linearly when the voltage increases linearly from 0 to +2 V. After the strip reaches its maximum deflection at +2 V, it must linearly bend back to finish a scan cycle. For this to occur, the mobile positive ions shifted to one side of the IPMC strip must be drawn to another side by the electrical field. So the applied voltage must switch to 0 V, change its electrical polarity, and increase linearly from 0 to −2 V. With the modified waveform, the scanning range depends not only on the voltage magnitude but also on the frequency. Generally, lower frequencies (as little as 0.1 Hz) or higher voltage (as much as 6 V) lead to higher displacement.[9–11] At a frequency of 1 Hz, the 30 mm × 2 mm IPMC cantilever can generate a stable linear scanning movement with a displacement range of ~3 mm at ~2 V (Fig. 3).

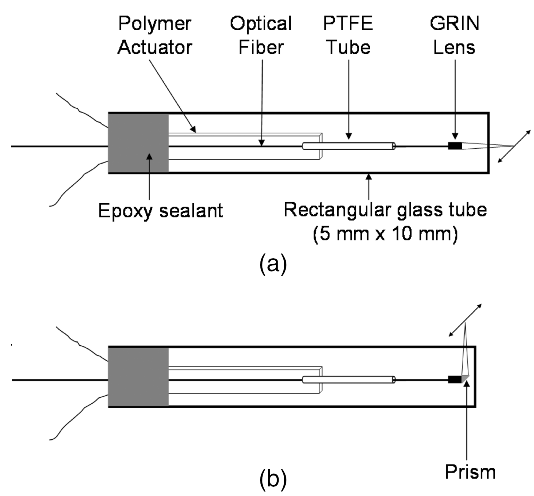

Fig. 2. Schematics of OCT probes incorporating the IPMC scanning cantilever: (a) forward scanning probe, (b) sidewise scanning probe.

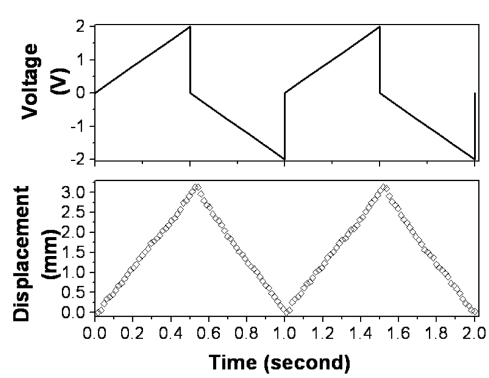

Fig. 3. Displacement of the fiber beam controlled by the IPMC scanning cantilever on a 2-V alternating waveform at 1 Hz.

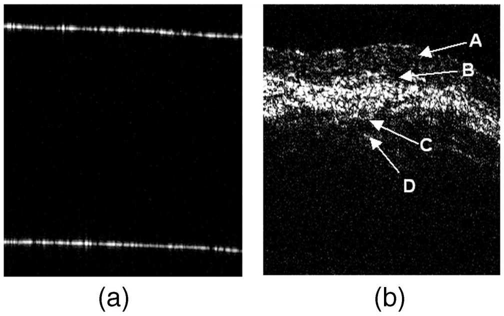

Fig. 4. Images obtained with OCT probes incorporating the IPMC scanning cantilever: (a) a clear plastic plate and (b) *in vivo* underside of a human tongue (A, nonkeratinized stratified squamous epithelium, B, basement membrane; C, salivary gland; D, lingual muscle fibers).

We used hand-held scanning probes for *in vitro* and *in vivo* OCT imaging at 2 V and 1 Hz. Figure 4(a) is an OCT image of a transparent plastic plate acquired with the forward scanning probe, which shows undistorted surfaces on both sides. Figure 4(b) is an *in vivo* OCT image of the underside of a human tongue, which we measured by softly touching the outer wall of the sidewise scanning probe against the tongue. Layered structures and detailed architectural morphology of the tongue can be visualized. Features such as a nonkeratinized stratified squamous epithelium, basement membrane, salivary gland, and lingual muscle layers can be clearly identified in the image and were confirmed by histology reference. The images in Fig. 4 have a transverse resolution of 20 μm; the lateral scan ranges are $\sim$3 mm, but the polymer actuator is able to produce larger scan ranges at higher driving voltage, for example, $\sim$6 mm at 5 V.

The IPMC cantilever can be cut by a scalpel into miniature strips of any desired size. Therefore a miniature IPMC strip can be integrated into a limited space in an OCT endoscope to image a variety of inner organs (esophagus, stomach, colon, bladder, etc.) in both forward and sidewise scanning modes. Although the scanning speed of the IPMC actuator is relatively slow (less than 10 Hz) compared with that of conventional actuators, its unique characteristics such as simplicity, ruggedness to handling, low cost, and low driving voltage make it ideally suitable for OCT and other biomedical systems.

In summary, we have constructed prototype hand-held OCT probes that can be incorporated into a polymer actuator and have demonstrated its feasibility for *in vitro* and *in vivo* OCT imaging. Linear movement of the polymer cantilever is achieved by application of a modified voltage waveform. The simple structure reduces the cost, and low driving voltage ensures safety for clinical applications. Two different optical designs produce both forward and sidewise scanning and make it possible to image the entire human oral cavity. The electroactive polymer cantilever has potential applications in OCT endoscopy for imaging of internal human tissues *in vivo*.

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