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Forward-viewing photoacoustic imaging probe with bundled ultra-thin hollow optical fibers

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Abstract

A photoacoustic imaging system composed of a flexible bundle of thin hollow-optical fibers is proposed for endoscopic diagnosis. In this system, a bundle of 127 hollow-optical fibers with an inner diameter of 100 μm was fabricated. The total diameter of the bundle was 2.1 mm, and the minimum bending radius was around 10 mm. Owing to the small numerical aperture of hollow optical fibers, a high resolution image was obtained without using a lens array at the distal end. In the imaging system, the hollow fibers in the bundle were aligned at the input end, so the hollow fibers were sequentially excited by linearly scanning the laser beam at the input end. Photoacoustic imaging systems consisting of the bundled fibers for excitation of acoustic wave and piezoelectric probes for detection of photoacoustic signals were built. By using the systems, photoacoustic images of blood vessels in the ovarian membrane of fish were taken to test the feasibility of the system. As a result, photoacoustic images of the vessel were successfully obtained with a laser fluence of around 6.6 mJ cm⁻².

Keywords: photoacoustic imaging, optical fiber probe, hollow optical fiber, endoscopic imaging

1. Introduction

Photoacoustic imaging is expected to become a new diagnostic tool specifically for observing blood capillaries that are peripherally generated with early cancer tissues [1] because the photoacoustic method offers larger penetration depths than optical imaging methods and higher resolutions than ultrasonic imaging [2, 3]. An endoscopic photoacoustic imaging system is also highly expected for non-invasive, diagnostic imaging of digestive tracts and cardiac blood vessels. Wang et al developed an endoscopic side-viewing probe that uses a rotating mirror for scanning an excitation laser beam in the radial direction and an ultrasonic transducer at the distal end of the probe for detecting the photo-acoustic signal. Their imaging probes were so thin as to be inserted into a working channel of endoscopes. In addition, they developed a hybrid imaging probe combining a photoacoustic device and an ultrasonic imaging probe. However, the uses of those probes are limited in side-view imaging, and they have a rigid part at the distal end because they have a scanning mechanism equipped at the end [4–6]. Another group reported on the development of an optical-fiber based, handheld photo-acoustic microscope. They used a bundle of optical fibers to obtain optical-resolution images; therefore, the system has no scanning mechanism at the distal end. However, the system still uses an ultrasonic transducer for detecting acoustic waves, and an endoscopic imaging system has not been developed [7, 8]. Beard et al developed an all-optical photoacoustic imaging system using a coherent fiber bundle with a polymer-film-based Fabry–Perot (FP) sensor placed at the distal end for detecting acoustic waves. They
combined the fiber bundle with a multimode optical fiber to obtain photoacoustic images without an ultrasonic transducer [9].

Our group proposed a thin and flexible all-optical photoacoustic imaging fiber-probe that consists of a bundle of hollow optical fibers with an inner diameter of 320 μm for delivering excitation laser beams and a single-mode optical fiber with thin polymer film at the distal end for acoustic detection [10]. Forward-viewing photoacoustic images were obtained using scanning excitation beams at the input end of the fiber bundle. No scanning mechanism was used at the distal end, so no rigid part was necessary. In addition, owing to the extremely small numerical aperture of hollow optical fibers, a parallel output beam was obtained; therefore, three dimensional images could be obtained. With this fiber probe, we succeeded in obtaining 3D images of blood vessel phantoms made of silicon tubing with a diameter of 1 mm. However, the imaging resolution that depends on the size of hollow optical fibers was limited to the submillimeter order, which is not good enough for imaging thin blood capillaries.

In this paper, we report our attempts to improve the resolution of an optical fiber probe by reducing the inner diameters of hollow optical fibers from 320 to 100 μm. In the photoacoustic imaging probe with hollow optical fibers, the resolution greatly depends on the fiber size because the divergence angle of the excitation beam emitted from the hollow optical fiber is very small. Reducing the fiber size should improve the flexibility, as well. Toward designing an endoscopic photoacoustic imaging system, we developed imaging probes using a bundle of ultra-thin hollow optical fibers and performed imaging experiments using biological samples.

2. Design and fabrication

Figure 1 shows a schematic of our photoacoustic imaging system that uses a bundle of hollow optical fibers to radiate an excitation laser beam on the sample. The input ends of the fiber are aligned, enabling beam scanning to be easily performed by using a linear-motion stage at the input end. As mentioned, a lens is not necessary at the output end because a nearly parallel beam is obtained from the hollow optical fibers, and this enables 3D photoacoustic imaging. A fiber-optic probe with a polymer film that functions as a FP interferometer attached at the distal end is utilized to detect excited photoacoustic waves, and this comprises the all-optical photoacoustic imaging system. In this paper, however, we focus on showing feasibility of the hollow-optical fiber bundle for excitation of photoacoustic wave without mechanical scanner at the distal end and common piezoelectric transducers were used for detection of acoustic waves.

We fabricated ultra-thin hollow optical fibers with an inner diameter of 100 μm by depositing silver thin film on the inside of glass capillaries using a liquid-phase deposition technique [11, 12]. A Q-switched, microchip Nd:YAG laser with high beam quality (Hamamatsu L11038-12, 532 nm wavelength, 1.2 ns pulse width, and 100 pps repetition rate) was used in the experiment as a light source for exciting photoacoustic waves to obtain a higher coupling efficiency to thin fibers. The laser beam was focused to a spot size of around 70 μm by using a lens with a focal length of 76 mm to excite the lowest order mode optimally in the hollow optical fibers [13]. The measured coupling loss was 1.8 dB for 100 μm core hollow optical fibers, and the transmission loss was 2.8 dB for the 27 cm long fiber.
First, we evaluated the lateral resolution of imaging systems based on hollow optical fibers. An almost parallel output beam was obtained from hollow optical fibers because of the extremely small numerical apertures (NA < 0.05). Therefore, assuming that the output beam has a Gaussian profile, beam radius \( w \) at distance \( z \) from the fiber’s output end is described as

\[
w = w_0 \sqrt{1 + \left( \frac{\lambda z}{\pi w_0^2} \right)^2},
\]

where \( \lambda \) is the wavelength, and \( w_0 \) is the beam radius at the output end that is 64% of the fiber radius for the lowest order mode.

In the experiment, we placed a tungsten wire with a diameter of 50 \( \mu \)m at a depth of 10 mm in pure water. The photoacoustic wave from a tungsten wire was detected by a single hollow optical fiber and the fiber placed on the water surface was moved at 50 \( \mu \)m step to obtain profiles. To detect acoustic waves, a PVDF needle hydrophone (Muller, Platte Needle Probe) was put inside the water. The diameter of the probe was 1.2 mm containing sensing part with a diameter of 0.5 mm and we found that the probe was almost omnidirectional. The bandwidth was 0.3–11 MHz and the sensitivity was around 1 mV/bar. Figure 2 shows the measured amplitude profile of the photoacoustic waves as a solid line and a Gaussian curve fitted to the measured data as a dotted line after applying normalization with maxima. From this fitted data, the measured lateral resolution was found to be 106 \( \mu \)m at FWHM, which coincides well with the beam diameter calculated using equation (1).

We also evaluated the imaging resolution in biomedical tissues using a 1%-intralipid solution having almost the same scattering coefficient as that of biomedical tissues \([2, 14, 15]\). Figure 3 shows the amplitude profile of the photoacoustic waves from a 50 \( \mu \)m diameter tungsten wire in the solution at depths of 200 and 400 \( \mu \)m. The profiles are normalized with the maximum amplitude of the one for 200 \( \mu \)m depth. These results show that the lateral resolution does not significantly change based on the depth, although the amplitude decreases due to absorption and scattering. From these results, we can expect that a photoacoustic imaging system based on hollow-optical fibers will be useful for visualizing 100 \( \mu \)m blood vessels in the submucosal layer.

3. Photoacoustic imaging

For photoacoustic imaging, we fabricated a bundle of 37 hollow-optical fibers with a diameter of 100 \( \mu \)m—for which the output end is shown in figure 4. The outer diameter of the bundle was 1.2 mm, and the minimum bending radius was around 10 mm. We used copper wires with a diameter of
150 μm as samples, and two wires were placed in an ‘X’ shape, as shown in figure 5(a). There was a gap of 700 μm between the wires in depth direction. The pulse energy of the excitation laser was around 100 μJ/pulse, and the acoustic signal was detected using the PVDF needle hydrophone.

Figure 5(b) shows a C-mode image of the wire placed in front of another one. This image was obtained by mapping the amplitude of photoacoustic signals from each fiber element onto closely packed hexagonal lattices after applying digital filtering by using MATLAB software. Figure 5(c) is a B-mode image that was calculated using envelope demodulation of the C-mode images. Figure 5(d) shows a three-dimensional image created by reconstructing superimposed B-mode images using Image J software. The image resolution was improved from that of our previous system by using the bundle of ultra-thin hollow optical fibers, and we succeeded in identifying two metal wires placed at different depths.

Next, we fabricated a fiber bundle with 127 hollow optical fibers with an inner diameter of 100 μm to improve the field of view for photoacoustic imaging. The output end of the bundle is shown in figure 6 and the area of the end surface was enlarged 3.5 times over that of the former bundle with 37 fibers. The outer diameter of the bundle was 2.1 mm, and the minimum bending radius was still as small as 10 mm. By using this bundle, we tried to obtain photoacoustic images of blood vessels in biological samples. In the experiment, a lead zirconate titannate (PZT) probe that has higher sensitivity and directivity than PVDF probes was used because the excitation laser power should be kept low not to cause a damage on the sample. The PZT probe with a sensing diameter of 7 mm had the bandwidth of 4–10 MHz and the sensitivity of around 100 V/bar. The probe was placed so as to showing highest sensitivity and preamplifier with a gain of 46 dB was used for signal detection.

Prior to photoacoustic imaging experiments, the positions of the output ends of each fiber element were observed and recorded by using a conventional CCD camera. The amplitude of the photoacoustic signals was set at the recorded positions so as to obtain images that exactly reflect the excitation patterns of the laser beam. In addition, to suppress the effect of variation of fibers’ transmission losses, we collected photoacoustic signals measured for samples beforehand using a reference amplitude pattern that was obtained for a uniform medium such as a colored plastic tape.

Figure 7 shows microscopic images and C-mode photoacoustic images of three different places of ovarian membrane of fish (cod roe). The blood vessels were clearly observed as shown in the figure and observation through a microscope showed that the thicknesses of the vessels varied from 160–600 μm. It was also seen that the changes in the photoacoustic amplitude reflect the differences in hemoglobin concentration of the vessels. In this experiment, the pulse energy of the radiated laser beam was as low as 0.5 μJ/pulse. This corresponds to the fluence of 6.6 mJ cm⁻², which was much lower than the American National Standards Institute safety limit (20 mJ cm⁻²) [16]. Therefore, we found that our system can be safely used in the photoacoustic imaging of biological tissues.

4. Conclusion

We developed a photoacoustic imaging system using a bundle of thin hollow optical fibers to improve the image resolution of forward-viewing photoacoustic probes. The diameter of each hollow fiber was 100 μm, and a measured lateral resolution of 106 μm was obtained owing to a parallel output beam from hollow optical fibers with an extremely small numerical aperture. Experiments conducted using an intralipid solution found that the resolution does not change substantially in highly diffusing biomedical tissues. We also performed three-dimensional imaging of copper wires of 150 μm diameter placed 700 μm deep by using a hollow fiber bundle with 37 elements. For photoacoustic imaging of biomedical samples, a fiber bundle with 127 elements was fabricated to improve the field of view of the probe. By using this bundle, we succeeded in photoacoustic imaging of thin blood vessels in the membrane.
of cod roe. By combining the hollow fiber bundle with an optical-fiber-based acoustic probe, an all-optical photoacoustic probe with high flexibility is expected for endoscopic forward-view imaging. We are currently working on improving the sensitivity of the fiber-based acoustic probe. The results of the combined system will be reported elsewhere.

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