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High-Sensitivity Fiber-Optic Ultrasound Sensors for Medical Imaging Applications

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Abstract

This paper presents several designs of high-sensitivity, compact fiber-optic ultrasound sensors that may be used for medical imaging applications. These sensors translate ultrasonic pulses into strains in single-mode optical fibers, which are measured with fiber-based laser interferometers at high precision. The sensors are simpler and less expensive to make than piezoelectric sensors, and are not susceptible to electromagnetic interference. It is possible to make focal sensors with these designs, and several schemes are discussed. Because of the minimum bending radius of optical fibers, the designs are suitable for single element sensors rather than for arrays.

Keywords

Electromagnetic interference; laser interferometer; optical fiber; ultrasound sensor

INTRODUCTION

In certain ultrasound imaging applications, such as inside magnetic resonance scanners[1] or in Hall effect imaging devices,[2] nonmetallic ultrasound sensors are preferred because of their immunity to electromagnetic interference. This paper describes the development of high-sensitivity fiber-optic ultrasound sensors suitable for such applications. These sensors are not affected by external electromagnetic fields. They are relatively simple to fabricate and have lower material cost than piezoelectric sensors.

In the past, fiber-optic ultrasound sensors have operated either by sensing pressure-induced refractive index changes in the fiber or surrounding medium,[3,4] or by monitoring pressure-induced displacements of a membrane structure on the tip of the fiber with high precision interferometers.[5,6] Due to the small diameter of optical fibers, these sensors are miniaturized to suit a wide range of applications. The displacement-based sensors have higher sensitivity and can detect pressure pulses of kPascal amplitudes in the MHz frequency range. The detection threshold is ultimately determined by the amplitude of displacement in the liquid surrounding the fiber tip relative to the laser wavelength.

For imaging applications, a higher sensitivity is desired. This can be achieved if the size of the sensor is not limited to the diameter of the fiber. The basic idea is derived from high sensitivity fiber-optic hydrophones in the audio frequency range (for a comprehensive introduction to optical fiber based acoustic sensors, see reference 7). A typical hydrophone consists of a thin optical fiber wound on a substrate. When immersed in water, acoustic pressure waves cause strains in the substrate and changes in the length of the optical fiber. The length change is measured at high precision by incorporating the fiber into the light path of one arm of a laser interferometer. High sensitivity is achieved from the short wavelength of the laser and the high gain produced by many windings of fiber in the hydrophone.

In extending this idea to ultrasound sensors, one must consider the much shorter acoustic wavelength. In optical fiber hydrophones, the entire sensor contracts or expands in response to the pressure changes in the surrounding medium. In ultrasound imaging, the acoustic wavelength in water is on the order of 1 mm or less, much lower than the minimum bending radius of the most flexible optical fibers. Thus, the size of the sensor is on the order of multiple wavelengths. The description of quasistatic strain is not valid, and wave propagation inside the sensor needs to be considered.

Besides sufficient sensitivity, medical imaging applications also require the sensor to be compact, rugged and able to focus to a specific depth. Based on these considerations, the following sensor designs were constructed and tested.

SENSOR DESIGNS AND THEORY OF SENSOR OPERATION

As stated above, the detection mechanism is that ultrasonic pressure waves cause strains in the optical fiber, thus modulating the phase of the light passing through the fiber. The fiber-based Michelson and Mach-Zehnder interferometers used to measure the strains are shown in figure 1. The sensing arm contains the segment wound around the sensor, while the reference arm contains a segment wound on a piezoceramic cylinder. Low-frequency thermal fluctuations and external perturbations cause phase drift between the two arms, leading to light intensity fluctuations at the output. These are compensated with a negative feedback circuit, providing voltage to the piezoceramic cylinder in the reference arm. The feedback circuit responds to audio and lower frequency fluctuations, so there is no signal loss in the ultrasound range. The gain and dc offset of the feedback voltage are adjusted to maintain a phase difference of 90° between the two arms. If the light intensity in each arm is I_0, any small strain ΔL induced by the ultrasonic waves results in a phase change $\Delta\varphi$ in the sensing arm and a change in the light output of the interferometer ΔI:

$$\Delta I=2I_0\sin\Delta\varphi=4\pi I_0\frac{\Delta L}{\lambda}, \tag{1}$$

where λ is the laser wavelength within the optical fiber, and ΔL is much smaller than λ.

Two different sensor designs are tested. In the first design,[8] a single-mode optical fiber is wound in a helix and glued to a thin flexible backing disk. The design is shown in figure 2. For testing, the sensor is immersed in a water tank. Ultrasonic waves produced in the tank may cause volumetric expansion and compression of the backing disk (the breathing mode), or they may cause it to wobble. Both forms of deformation change the strain in the fiber disk, which is detected by the interferometer.

A limitation of the planar disk geometry is that glass-core optical fibers have minimum bending diameters of 5 mm or larger. In order to keep the thickness of the fiber disk small compared to the acoustic wavelength, only one or two layers of fiber can be used. To wind a sufficient length of fiber to improve sensitivity, the diameter of the disk needs to be 25 mm or larger. This is larger than most single element probes used in medical ultrasound. The thin flexible backing disk may also change shape when pressed against the sample, thus changing the acoustic profile of the sensor.

The second design overcomes these limitations by manipulating the ultrasound wave-front. In this design, the optical fiber is wound into a cylinder. Placed inside the cylinder is a coaxial conical reflector, with the tip of the cone facing the incoming ultrasonic wave (Fig. 3). When immersed in the test tank, incoming waves propagating parallel to the axis of the cylinder are reflected radially outward toward the cylindrical surface of the fiber spool. If the angle of the cone is 45°, an incoming planar wavefront is reflected into a cylindrically-outgoing wavefront, and impacts the fiber cylinder simultaneously. The overall length change in the fiber is the sum of the changes in all the turns of the cylinder, which magnifies the signal many fold. This geometry also allows the sensor diameter to be as small as 5 mm, the minimum bending diameter of the fiber.

In a more robust construction, the fiber is wound around a solid cylindrical form with a cone-shaped hollowing at one end (Fig. 4). The form is made of a plastic material with acoustic impedance similar to water. The plastic-air interface of the cone serves as the reflector. The reflection is nearly complete since the acoustic impedance of air is orders of magnitude lower than that of solid material. To receive an ultrasound signal from an object such as tissue, all that is necessary is to make a good acoustic contact between the object and the flat end of the form. Compared to the immersion design in figure 3, there is some sensitivity loss from the acoustic impedance mismatch between the plastic material and water. This loss can be recovered with impedance matching layer(s).

The sensitivity of the cylindrical sensors can be estimated from the relationship between the ultrasound pressure and changes in the radius of the fiber cylinder. Consider a plane wave of pressure P incident on the sensor. Denote the acoustic impedance of the cylinder as Z, the number of turns of the fiber as N, the linear Young's modulus of the fiber as Y (force-strain ratio along the fiber), the laser wavelength in the fiber as λ, the radius of the cylinder as R, the pressure-induced radius change as ΔR, and the stretch in the fiber length as ΔL. The incident plane wave is reflected by the cone and propagates radially to the fiber cylinder (Fig. 3). Because of this process, the pressure on the cylinder decreases from P at the base level of the cone to near zero at the tip level of the cone. For approximate estimates, the average pressure on the cylinder is taken as $P/2$. To obtain the change of radius ΔR under this pressure, one needs to include the constrictive pressure on the cylinder by the expansion of the fiber spool. This pressure is

$$P_f = N \Delta R Y / R^3. \tag{2}$$

The incident pressure on the cylinder wall will approximately balance the sum of this and the internal stress associated with the radial strain in the cylinder:

$$P/2 = P_f + \Delta R Z \omega. \tag{3}$$

Substituting Eq. (2) into Eq. (3), ΔR can be expressed as

$$\Delta R = \frac{P/2}{Z\omega + NY/R^3}. \tag{4}$$

The laser phase change can be expressed as

$$\Delta\varphi=\frac{2\pi}{\lambda}\Delta L=\frac{2\pi}{\lambda}N2\pi\Delta R=N\frac{2\pi}{\lambda}\frac{P}{z\omega+NY/R^3}. \tag{5}$$

It should be noted that all the fiber-optic sensors described here are displacement based—displacements in the medium directly translate into fiber length changes. As expressed in Eq. (1), this results in a proportional change in the light output of the laser interferometer. Because displacements are proportional to pressure divided by frequency, the sensitivity of the sensors in radians per unit pressure decreases with frequency. This is seen in the measurements described below. Other mechanical and optical factors that may affect the performance of the sensors at higher frequencies will be described later in the Discussion section.

METHODS AND RESULTS

Sensors were made from Corning single-mode 80 μm diameter fiber. In the first design, the fiber is wound into a disk of 10 mm inner diameter and 25 mm outer diameter, and the fiber disk is glued to a 1 mm thick polyethylene disk of 26 mm diameter (Fig. 2). This sensor is incorporated into one arm of a Michelson interferometer (Fig. 1). Thermal fluctuations and external vibrations are compensated by winding the other arm of the interferometer onto a piezoceramic cylinder, and applying on the piezoceramic cylinder a feedback voltage from the output light intensity. The feedback voltage is low-pass filtered and only compensates for fluctuations in the audio range and below, where most of the thermal drifts and environmental noise occur. The light source of the interferometer is a 1.3 μm wavelength high-coherence solid-state laser (Lightwave Electronics). This sensor was tested in a water tank where ultrasonic pulses were generated with broadband piezoelectric transmitters (Krautkramer-Branson Alpha immersion transducers), shown in figure 5. In order to quantify the sensitivity, the acoustic pressure at the location of the fiber sensor needs to be measured. This was done by repeating the measurements with piezoelectric transducers of known sensitivities. The sensors and transmitters are directional and nonfocused, and it was found that direction alignment between the transmitter and the receiving sensor was crucial for reliable measurements, especially at higher frequencies.

Four sets of measurements were performed with broadband transmitters of center frequencies 1 MHz, 2.25 MHz, 3.5 MHz and 5 MHz, respectively. These measurements overlapped in the frequency range of 1 to 4 MHz; thus, each frequency is covered by two to four measurements. Figure 6 shows the average sensitivity and standard deviation of the measurements over the entire frequency range. The lowest two frequencies were measured only with the 1 MHz transmitter. Since the fiber sensor directly measures displacements rather than pressure, its sensitivity with respect to acoustic pressure decreases with frequency. The noise of the entire detection system was dominated by the intensity fluctuations of the laser source, and was on the order of 0.05 mRadian/√MHz in the 1 MHz to 4 MHz range. This noise level sets the detection threshold at about 5 Pascal/√MHz at 1 MHz, and about 10 Pascal/√MHz at 4 MHz.

The second design was realized in two sensors of different constructions. In the first construction, the optical fiber is wound around a thin polyethylene cylinder, centered around a conical aluminum reflector (Fig. 3). The fiber cylinder consisted of 30 turns in one layer. The overall diameter of this sensor is 13 mm, and strain in the fiber is measured with a Mach-Zehnder interferometer (Fig. 1). The sensitivity was measured as described for the first design. The result is shown in figure 6. The detection threshold of the overall system is about 3 Pascal/√MHz at 1 MHz and 15 Pascal/√MHz at 4 MHz. Compared to the fiber disk design, there is no loss of sensitivity.

The sensitivity of this sensor can also be estimated with Eq. (5). The nominal acoustic impedance of the cylinder is 2.0 Mrayl, and the tensile Young's modulus of the fiber is estimated from the bulk Young's modulus of glass to be 300 N. With these parameters the calculated sensitivity at 1 MHz is 45 μRadian/Pascal, and at 4 MHz is 11 μRadian/Pascal. These numbers are about twice the measurements (Fig. 6), possibly due to acoustic reflections at the water-polyethylene interface and the polyethylene-fiber interface.

In the second construction, optical fiber was wound on a solid cylindrical form with a cone shaped hollowing at one end and a flat surface at the other (Fig. 4). The form was made of plexiglass and had a 13 mm diameter (Fig. 7). Figure 6 shows the sensitivity-frequency relation. The sensitivity is reduced by about 50% from the water-filled version, likely due to acoustic mismatch between plexiglass and water.

The dynamic range of these sensors can be determined from figure 6 and Eq. (1). For the Michelson and Mach-Zehnder interferometers, the output intensity is approximately linear to the laser phase shift up to 1 radian from the quadrature point. This corresponds to an acoustic pressure of up to 40 kPa at 1 MHz, and 200 kPa at 4 MHz.

CONCLUSION AND DISCUSSION

We have described several fiber-optic ultrasound sensors based on measuring pressure-induced strains in fiber spools. According to Eq. (5), the sensitivity of these sensors scales with the number of turns in the fiber spool, but only up to certain limiting points. One limiting factor is the longitudinal tension within the fiber. As shown in Eq. (5), when this tension becomes the main resistance toward changes in the radius of the sensor, the sensitivity does not increase further with more turns of winding. Another limiting point occurs when the total thickness of the fiber layers approaches half the acoustic wavelength. Beyond this thickness strains in successive fiber windings reverse sign. The third limiting point occurs when the time for light to traverse the entire length of the fiber approaches the period of the ultrasound signal. Beyond this point, the pressure-induced phase shift in the light will not constructively sum. For a sensor of 15 mm diameter and operating at 4 MHz, the first limiting point is reached at approximately 100 turns, the second at 60 turns (2 layers of 30 turns each), the third at approximately 1,000 turns. The tension and thickness constraints are therefore most relevant. The latter is dependent on the acoustic wavelength in the fiber layers. At ultrasound frequencies exceeding 10 MHz, the acoustic wavelength approaches the diameter of the optical fiber, and the basic premise of this type of sensor design is not valid. The sensitivity above 10 MHz is expected to fall off quickly. These two limiting points increase with the radius. Correspondingly the achievable sensitivity will be higher for larger diameter sensors.

With the simple Michelson and Mach-Zehnder interferometers described above, the detection thresholds of the fiber-optic sensors are approximately 5 Pascal at 3 MHz and below. This is about 50 times the detection threshold of piezoelectric (PZT5H or pvdf based) sensors of similar sizes. This threshold is determined by the intensity-noise of the laser source. Therefore with differential light detection techniques the sensitivity of these fiber-optic sensors may be increased tenfold or more. Other interferometers with higher finesse values should also improve the sensitivity.

The measurements above showed that the immersion sensor of the second design had the highest sensitivity, while the planar disk design had the broadest bandwidth. However, neither design is as practical as the hollow-cone sensor for imaging applications. The planar disk sensor is large and lacks rigidity, while the immersion sensor needs to be filled with water and sealed with a thin membrane. The hollow cone sensor is of rigid construction and only the flat end of the sensor needs to be in contact with the subject. It can be used in much the same way as conventional piezoceramic transducers. Although the acoustic impedance mismatch between plexiglass and water decreases its sensitivity, this can be recovered with matching layers. For these reasons, the hollow cone design is the most promising one for practical applications.

Imaging sensors are often focused to improve resolution. Acoustic focusing can be realized in many ways in the fiber-optic sensors. In the second design involving the reflector cone, the shape and angle of the cone relative to the cylindrical axis determine the location of the focal point. Alternatively, a curved front surface can be used. Another option is to use a cylinder that expands slightly toward the front. Which is the optimal scheme will be decided by further experiments.

In summary, for imaging applications in environments where strong electromagnetic interference exists, such as Hall effect imaging or ultrasound imaging in combination with magnetic resonance scans, fiber-optic sensors may be the optimal choice. The above described sensors are receive-only devices, and therefore need to be combined with pulsing elements to function in echo-based imaging. Because optical fibers generally have minimum bending diameters, these designs are not suitable for making compact arrays or small intracavity probes. However it is conceivable that other interferometer-based designs such as miniature Fabry-Perot sensors may fill these roles.

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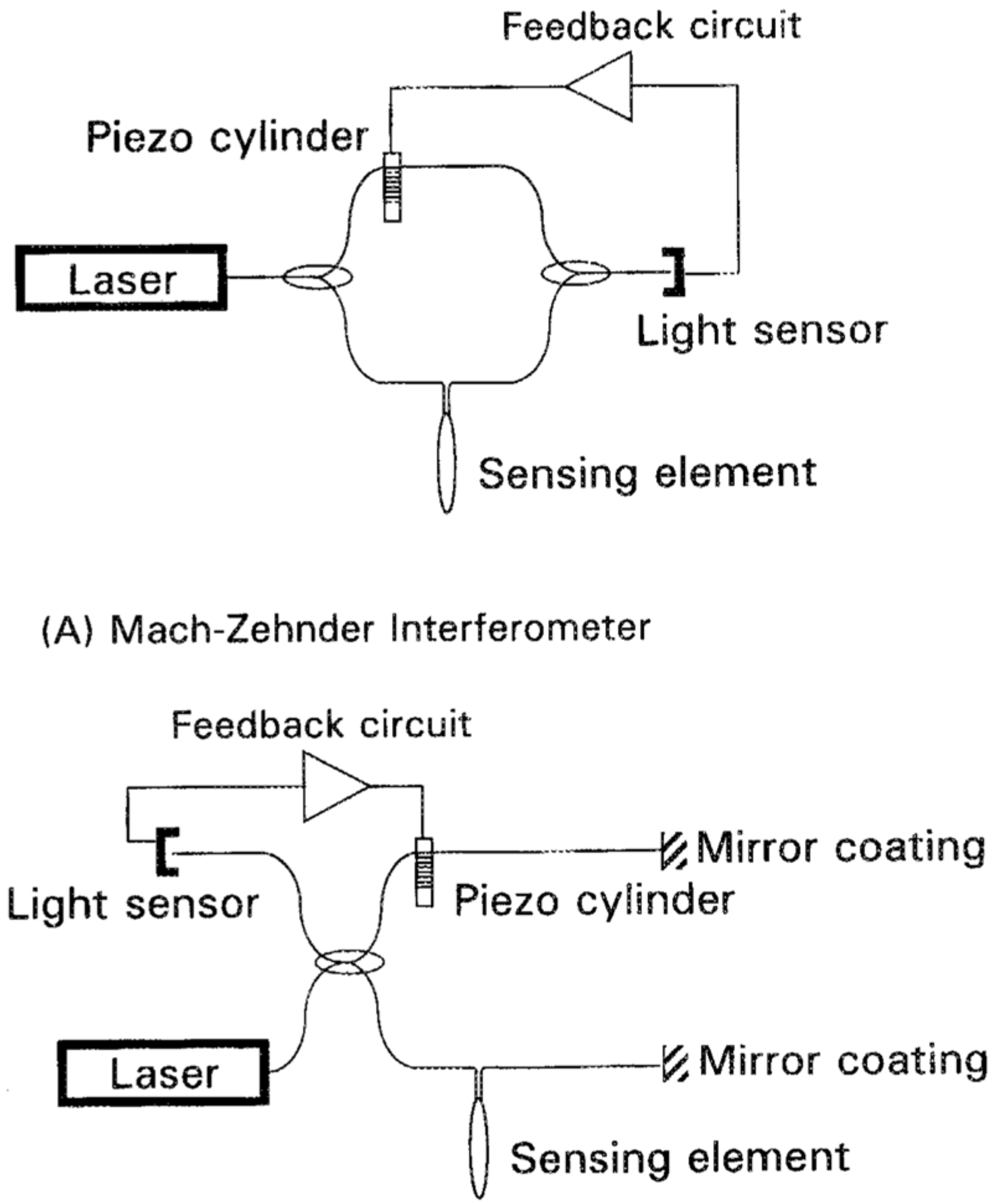

FIG. 1.

Fiberized laser interferometers used to measure the ultrasound-induced strain in the fiber-optic sensors. (A) Mach-Zehnder interferometer. The high coherence light from the laser is split into the reference arm and the sensor arm, then recombined. The combined light intensity is measured with a photodetector. The feedback voltage applied on the piezoelectric cylinder compensates for audio and lower frequency fluctuations in the lengths of the fibers. (B) Michelson interferometer. The laser is split into the reference arm and sensor arm, and reflected at the end of the fibers. The reflected light recombines and the light sensor measures the combined light intensity. The feed-back circuit and piezoelectric cylinder function the same way as in (A). In the Michelson interferometer, light passes through the sensor twice.

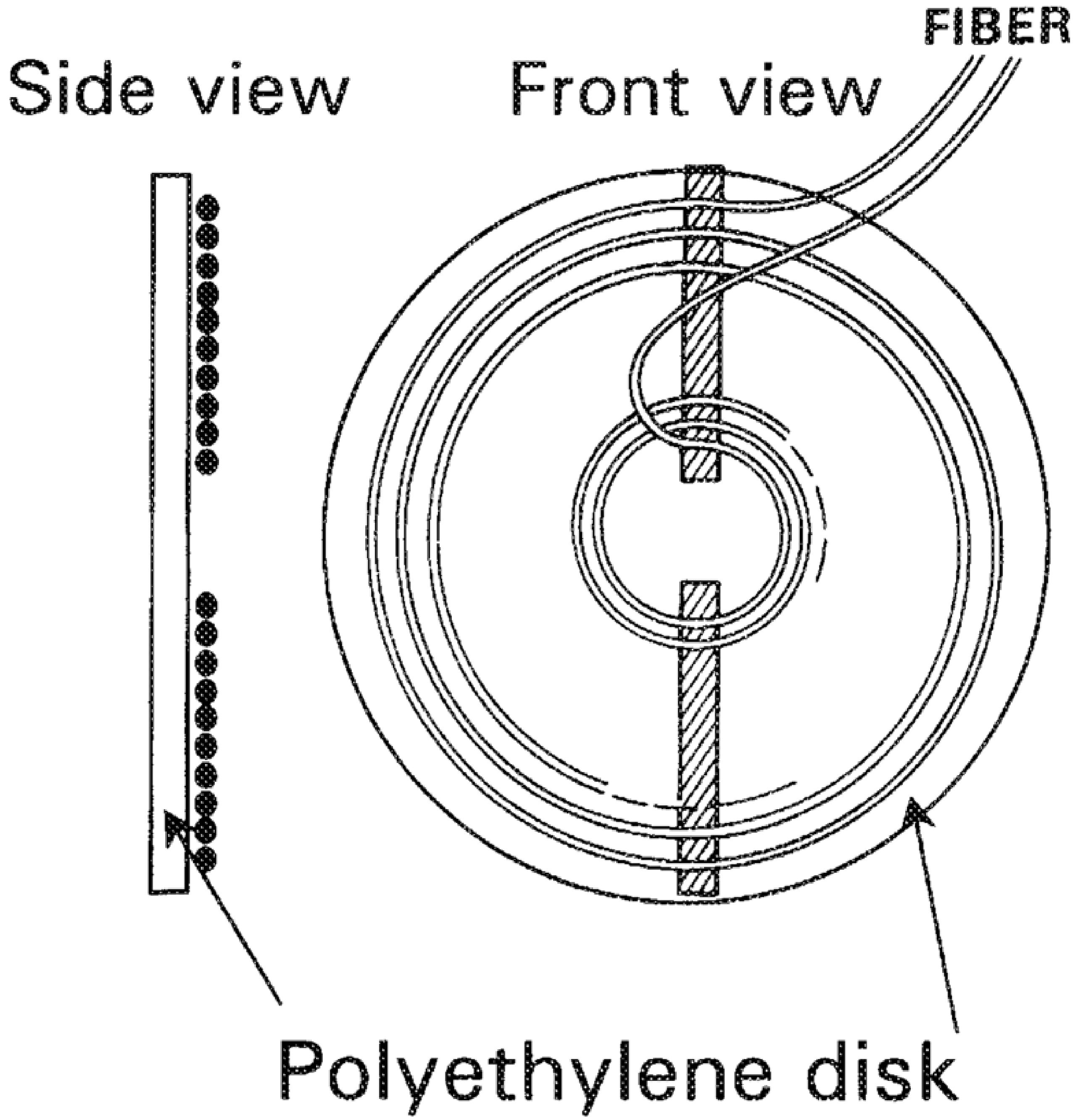

FIG. 2.

Design of first sensor. The optical fiber is wound into a planar disk and glued to a backing disk of polyethylene.

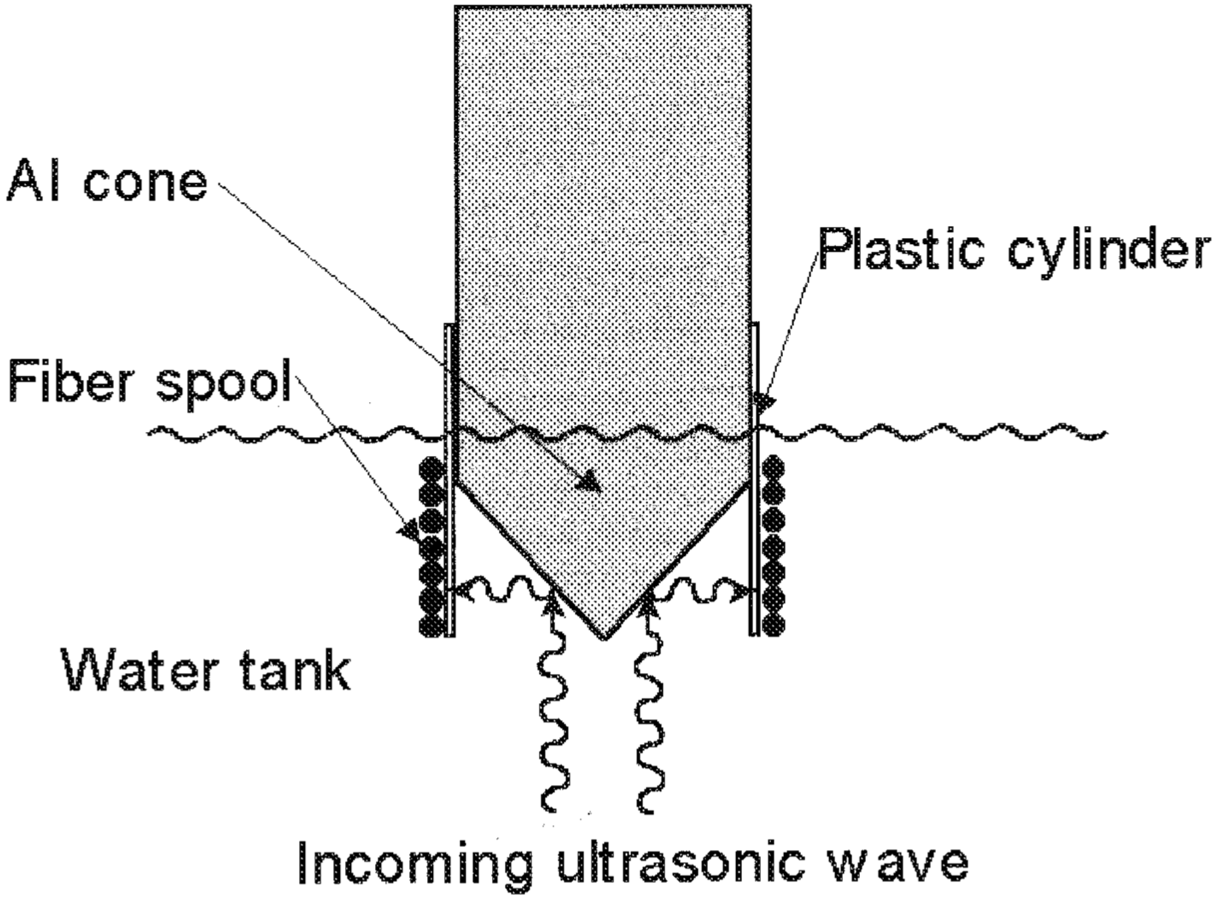

FIG. 3.
Design of second sensor. The optical fiber is wound on a thin polyethylene cylinder. A coaxial aluminum cone ultrasound reflector is inserted into the cylinder to direct the ultrasonic wavefront toward the fiber cylinder.

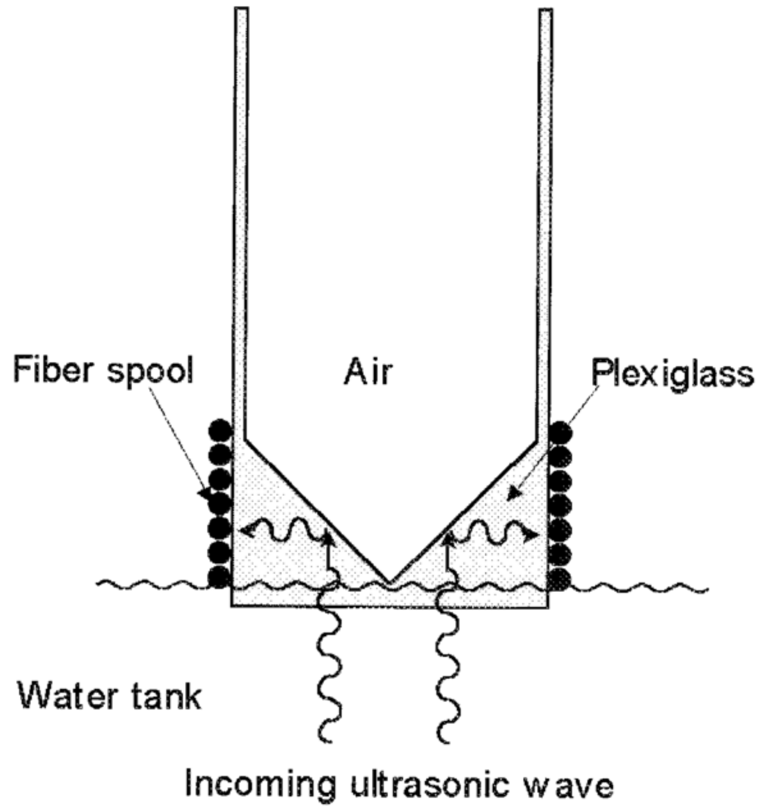

FIG. 4.
Practical construction of second design. The optical fiber is wound on a plexiglass cylinder with a conical hollowing in the back as the ultrasound reflector.

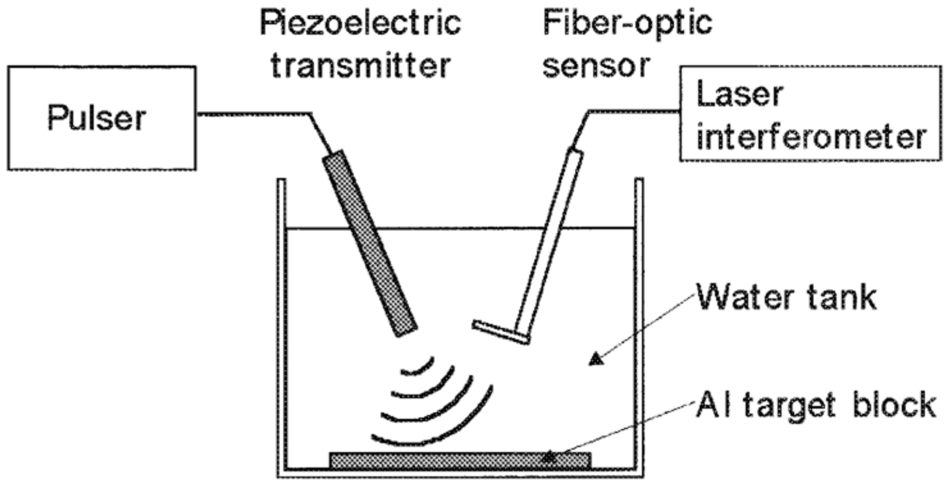

FIG. 5.
Arrangement for measuring sensitivity of the fiber-optic sensors. Ultrasonic pulses are emitted from the transmitter, reflected by an aluminum target block and received by the sensors.

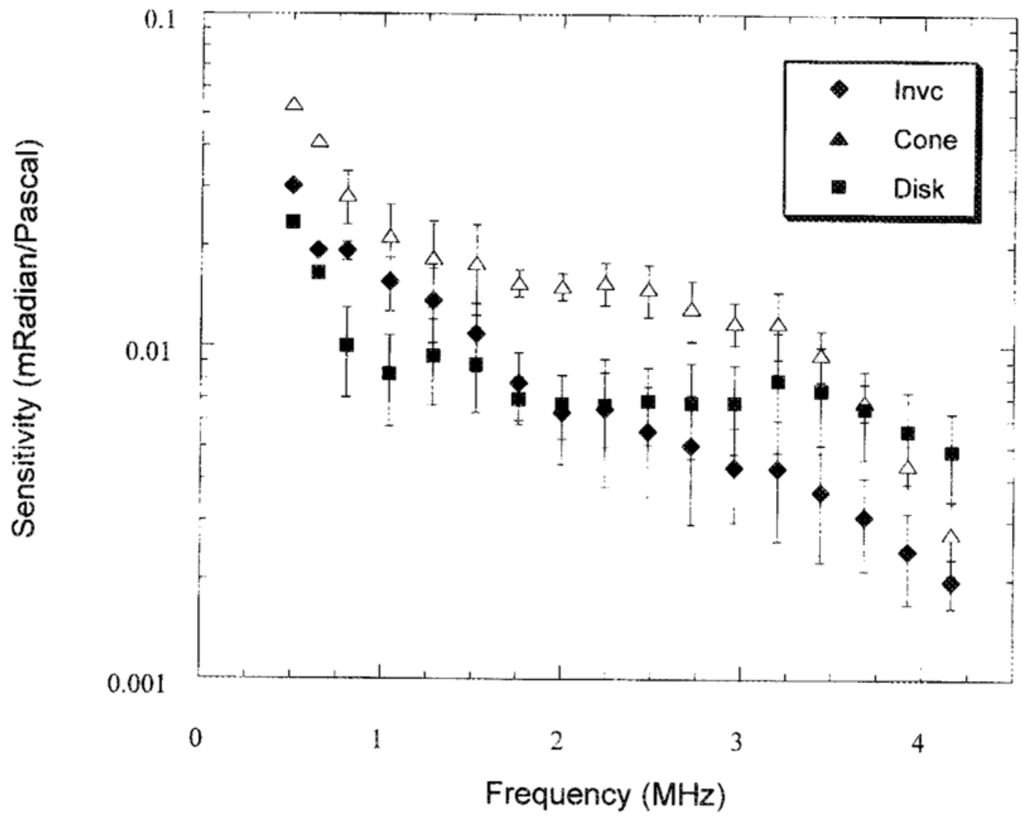

FIG. 6.
Sensitivity of the fiber-optic sensors from 500 kHz to 4 MHz, measured in milliradians per pascal. The first sensor was tested with a Michelson interferometer, which doubles its sensitivity when compared to a Mach-Zehnder interferometer. For this reason, the displayed sensitivity of the first sensor is half of the measured value. ■: the first sensor design, planar disk geometry. Δ: the second sensor design with aluminum cone reflector. ◆: the second sensor design with plexiglass cylinder and hollow-cone reflector.

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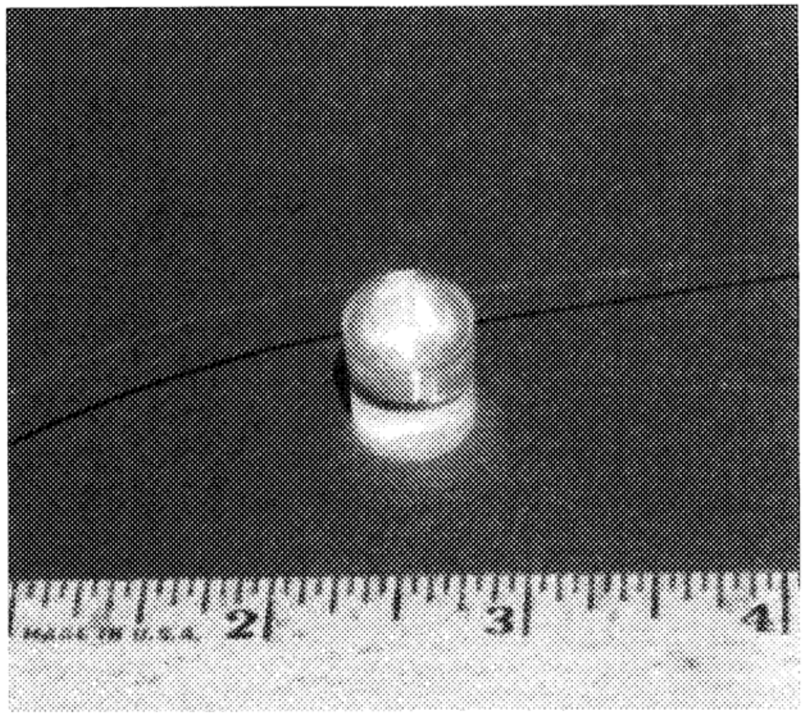

FIG. 7.

Photograph of the fiber-optic ultrasound sensor with the plexiglass form and hollow cone reflector. The ruler below is graduated in 1/16′ (1.5 mm).