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Design Considerations and Performance of MEMS

Acoustoelectric Ultrasound Detectors

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Abstract

Most single-element hydrophones depend on a piezoelectric material that converts pressure changes to electricity. These devices, however, can be expensive, susceptible to damage at high pressure, and/or have limited bandwidth and sensitivity. We have previously described the acoustoelectric (AE) hydrophone as an inexpensive alternative for mapping an ultrasound beam and monitoring acoustic exposure. The device exploits the AE effect, an interaction between electrical current flowing through a material and a propagating pressure wave. Previous designs required imprecise fabrication methods using common laboratory supplies, making it difficult to control basic features such as shape and size. This study describes a different approach based on microelectromechanical systems (MEMS) processing that allows for much finer control of several design features. In an effort to improve the performance of the AE hydrophone, we combine simulations with bench-top testing to evaluate key design features, such as thickness, shape, and conductivity of the active and passive elements. The devices were evaluated in terms of sensitivity, frequency response, and accuracy for reproducing the beam pattern. Our simulations and experimental results both indicated that designs using a combination of indium tin oxide (ITO) for the active element and gold for the passive electrodes (conductivity ratio = ~20) produced the best result for mapping the beam of a 2.25-MHz ultrasound transducer. Also, the AE hydrophone with a rectangular dumbbell configuration achieved a better beam pattern than other shape configurations. Lateral and axial resolutions were consistent with images generated from a commercial capsule hydrophone. Sensitivity of the best-performing device was 1.52 nV/Pa at 500 kPa using a bias voltage of 20 V. We expect a thicker AE hydrophone closer to half the acoustic wavelength to produce even better sensitivity, while maintaining high spectral bandwidth for

characterizing medical ultrasound transducers. AE ultrasound detectors may also be useful for monitoring acoustic exposure during therapy or as receivers for photoacoustic imaging.

I. Introduction

In applications ranging from ultrasound imaging to focused ultrasound therapy, knowledge of the acoustic field is critical. Traditional acoustic measuring devices such as fiber optic, membrane, and needle hydrophones are options for this purpose, depending on the application. These devices, however, are limited by certain features, such as sensitivity, acoustic bandwidth, cost of fabrication, and susceptibility to damage under high pressure [1], [2]. Membrane and needle hydrophones, for example, are very expensive and fragile. Fiber optic hydrophones are less expensive, but generally have lower sensitivity. Most of these devices also require a piezoresistive or piezoelectric material. This study examines an alternative type of acoustoelectric (aE) hydrophone that potentially complements or overcomes the limitations of standard devices.

Piezoresistivity refers to the pressure-induced stress that leads to a change in geometry of a solid material and a redistribution of its electrical charge [3]. Piezoresistive strain gauges, for example, have been used to detect the acoustic radiation force associated with an ultrasound beam [4]–[8]. Although the acoustic power was proportional to the electrical output of the strain gauge, these devices were unable to capture the phase of the acoustic wave. For semiconductors, the resistivity is determined by the electron charge, mobility, and density of charge carriers. The average atomic spacing in a semiconductor lattice changes under applied deformation or strain. A piezoresistive strain sensor of indium tin oxide (ITO) was used to measure changes in resistivity, but this device also failed to show how to capture the high-frequency modulation of an ultrasound beam [9].

AE theory, in a similar fashion, refers to the interaction and modulation of charged particles (e.g., electrons) in a material (especially solids and liquids) caused by stress waves associated with a propagating acoustic signal [10]–[12]. This wave–particle drag phenomenon couples electrical and mechanical energy and can be used to detect an acoustic wave. As ultrasound propagates through a conducting medium, the AE interaction signal is detected across two electrodes according to ohm's law and the aE effect. Other reported biological applications of the aE effect range from impedance imaging to mapping current flow in the heart [13]–[16]. It should be noted that piezo-resistive theory is similar to aE theory, although they have been historically used for different applications. Whereas piezoresistivity describes a linear change in the resistivity (ρ/ρ_0) of a material as a function of stress (σ) $\rho/\rho_0 = \pi^{\sigma_1} \sigma$ [9], the AE effect instead characterizes the change of resistivity with pressure $\rho/\rho_0 = -K_1 P$ [14]. The AE effect is more general (e.g., applies also to liquids) than piezoresistivity, because stress is usually reserved for solid materials, such as metals and semiconductors. Note that for isotropic solids ($\sigma = -P$), interaction constant K_1 is equal to the longitudinal piezoresistive coefficient $\sim \sigma_1$, and the AE effect explains the same phenomenon as piezo-resistivity.

We have previously described a type of inexpensive hydrophone that exploits the AE effect [17]. The AE hydrophone has attractive attributes not typically observed in other

hydrophones: simple construction, low cost, high sensitivity, resistance to damage from high-intensity ultrasound fields, and potentially wide bandwidth with an improved design. The active area may be composed of any type of resistor, such as graphite, a thick-film ceramic, or ionic gel. Initial designs, however, were difficult to fabricate with precision. Crude laboratory methods were employed, and fine control of the geometry of the active and passive areas was not feasible. This study implements both simulation and bench-top experiments to explore and help optimize design parameters of the AE hydrophone and facilitate rapid fabrication using MEMS technology for fine control of the size and electrical properties of the device.

II. Theory

The AE hydrophone exploits a small region of high current density $\mathbf{J}^I$ [the sensitivity zone (SZ)] to map an ultra-sound beam. The basic elements and geometry of one type of AE hydrophone are displayed in Fig. 1 [18]. The hydrophone and transducer coordinate systems are used to describe the electric field and acoustic pressure field. The voltage V^{AE} measured by a detector at coordinate x_0, y_0, z_0 in the electric field with a distributed current source $\mathbf{J}^I = \mathbf{J}^I(x, y, z)$ can be expressed in three dimensions using reciprocal theory [19], [20], given the assumption of farfield detection of the AE signal [21]. For a high-frequency (megahertz) AE signal V^{AE} , we expand the ultrasound pressure factor P or p into its subcomponents, such that

$$p(x, y, z, t) = P_0 b(x, y, z) a\left(t - \frac{z}{c}\right), \quad (1)$$

with ultrasound beam pattern $b(x, y, z)$ defined with respect to the transducer at the origin, pressure pulse amplitude P_0 , pulse waveform $a(t)$, and speed of sound c . The ultrasound wave field $p(x, y, z, t)$ can be simulated using Field II software [22], [23].

In Fig. 1(b), the center of the AE hydrophone is at $H(x_0, y_0, z_0)$, and the center of the transducer is at $C(0, 0, 0)$. Any point P in the ultrasound pressure field (x, y, z) can be described in the hydrophone coordinate system as $(x - x_0, y - y_0, z - z_0)$ or $\overrightarrow{HP} = \overrightarrow{CP} - \overrightarrow{CH}$ [17], [21].

$$V^{AE}(x_0, y_0, z_0, t) = -P_0 \iiint K_I \rho_0 (\mathbf{J}^L \cdot \mathbf{J}^I) (x - x_0, y - y_0, z - z_0) \left[b(x, y, z) a\left(t - \frac{z}{c}\right) \right] dx dy dz, \quad (2)$$

where K is the interaction constant (e.g., $\sim 10^{-10} \text{ Pa}^{-1} \text{ I}$ in ITO [24]), ρ_0 is the direct current resistivity, and $\mathbf{J}^I(x, y, z)$ is the electric lead field resulting from unit reciprocal current formed by the hydrophone's two sensing electrodes.

Because $K_I = K_I(x, y, z)$ and $\rho_0 = \rho_0(x, y, z)$ both depend on the material properties, they can be combined together with $\mathbf{J}^I(x, y, z)$, so that

$$w(x, y, z) = K_I(x, y, z) \rho_0(x, y, z) (\mathbf{J}^L \cdot \mathbf{J}^I)(x, y, z). \quad (3)$$

$V^{AE}(x_0, y_0, z_0, t)$ involves three convolutions of $w(x, y, z)$ and $p(x, y, z, t)$, so it can be calculated quickly by 3-D Fourier transform to accelerate the computing of convolutions:

$$\begin{aligned} V^{AE}(x_0, y_0, z_0, t) &= - \int \int \int w(x - x_0, y - y_0, z - z_0) p(x, y, z, t) dx dy dz \\ &= - F_{x_0, y_0, z_0}^{-1} \left\{ W(k_x, k_y, k_z) F_{k_x, k_y, k_z} [p(x, y, z, t)] \right\}, \end{aligned}$$

where F_{x_0, y_0, z_0}^{-1} is the 3-D inverse Fourier transform over (k_x, k_y, k_z) , W is the 3-D Fourier transform of $w(x, y, z)$, F_{k_x, k_y, k_z} is the 3-D Fourier transform over (x, y, z) .

According to (4), the pressure field $p(x, y, z, t)$ can be approximated from the measured $V^{AE}(x_0, y_0, z_0, t)$ and simulated $w(x, y, z)$ on the AE hydrophone.

$$p(x_0, y_0, z_0, t) = - F_{x, y, z}^{-1} \left\{ \left(\frac{F_{k_x, k_y, k_z} [V^{AE}(x_0, y_0, z_0, t)]}{W(k_x, k_y, k_z)} \right) \right\}, \quad (5)$$

where F_{k_x, k_y, k_z} is the 3-D Fourier transform over (x_0, y_0, z_0) , $F_{x, y, z}^{-1}$ is the 3-D inverse Fourier transform over (k_x, k_y, k_z) . In the simulation of $w(x, y, z)$, the physical properties of the hydrophone should match well with the actual device to reconstruct the original ultrasound field. These include the hydrophone geometry (shape, width, and thickness) and conductivity ratio between the conductive area (passive element) and sensitivity zone (active element).

When the shape of the conductive electrodes is a bowtie or dumbbell, $\mathbf{J}^L$ approximates a dimensionless delta function $\delta(x, y, z)$ at the sensitivity zone, such that $|\mathbf{J}^L| \approx J_0 \delta(x, y, z)$, where J_0 has units of m^{-2} because of the unit reciprocal current. For a two-electrode AE hydrophone, the injecting electrodes are the same as the detecting electrodes, but not for a four-electrode AE hydrophone. In both cases, $\mathbf{J}^I = I \mathbf{J}^L$ with I denoting the injected current [20]. Assuming far-field detection of V^{AE} , the following simplifications can be made for modeling:

$$w(x, y, z) = K_1 \rho_0 I |\mathbf{J}^L|^2 \approx K_1 \rho_0 I J_0^2 \delta(x, y, z). \quad (6)$$

V^{AE} can then be expressed as

$$V^{AE}(x_0, y_0, z_0, t) \approx - \int \int \int K_1 \rho_0 I J_0^2 \delta(x - x_0, y - y_0, z - z_0) \times p(x, y, z, t) dx dy dz \quad (7)$$

where h has units of meters and depends on the ratio of the cross-sectional area of the sensitivity zone (SZ) over the length. Eq. (7) assumes the SZ is modeled as an ideal spatial delta function and simply shows a linear relation between the output voltage and the acoustic pressure. Therefore, the pressure field is proportional to the measured V^{AE} and is approximated by

$$p(x_0, y_0, z_0, t) \approx \frac{-h}{K_1 \rho_0 I} V^{AE}(x_0, y_0, z_0, t). \quad (8)$$

III. Methods

A variety of AE hydrophones were designed, constructed, simulated, and tested based on a structure with a small resistive element (SZ) and connecting electrodes. The electrodes had the same or higher conductivity than the SZ. We hypothesized that the performance of the hydrophone can be improved by increasing the conductivity ratio σ_r , defined as the ratio between the electrode conductivity and that of the sensitivity zone. When the SZ is composed of the same material as the electrodes, $\sigma_r = 1$. Larger σ_r indicates greater electrode conductivity than the SZ.

A. Fabrication of AE Ultrasound Detector

The AE hydrophones were fabricated in a clean room environment [18]. Glass substrates were coated with ITO and the hydrophone pattern was created in the ITO by a positive photolithographic process. Half of the samples were left with the pattern (bowtie or dumbbell) unaltered in ITO; gold electrodes were applied to the other half, up to the boundaries of the SZ. The gold, with a thin titanium layer to provide adhesion to the substrate, was applied by electron-beam deposition and patterned with a photolithographic process similar to that used on the ITO, using fiducial markers to co-register the pattern. The gold etching was done with a potassium iodide etchant, and the titanium was etched with a 1:2 ratio of ammonium hydroxide and hydrogen peroxide. A small area was left unmasked in the center of each device during the exposure step of the electrode formation process to create sensitivity zones of finite resistance in ITO between the gold electrodes. After the photolithographic process was completed, the glass substrates were diced and spin-coated with Cytop (Asahi Glass Co. Ltd., Tokyo, Japan) for waterproofing and insulation, leaving a small area uncoated for connection to the electrode leads. Once the leads were attached, the connections were waterproofed with UV-cured epoxy.

Four primary factors were investigated during the final design of the AE hydrophone: shape, width, thickness, and conductivity ratio. Three different shapes: triangular bowtie, circular, and rectangular dumbbells determine the current distribution in the lead field. Several AE hydrophones were created on glass substrates, either with or without gold electrodes, using two design patterns: dumbbell and bowtie. The dumbbell and bowtie in 100% ITO (5×10^{-4} to $5 \times 10^{-5} \Omega\cdot\text{cm}$) have SZs of $75 \mu\text{m}$ and $200 \mu\text{m}$ square. For the Au/ITO bowtie hydrophone, ITO fills the center area, approximately 1 mm square, and gold ($2.4 \times 10^{-6} \Omega\cdot\text{cm}$) covers the surrounding area; the conductivity ratio of Au/ITO is in the range of 20 to 100.

B. Experimental Setup

In Figs. 1(a) and 2(a), one pair of electrodes was used to inject 200 Hz alternating current and another pair was used for detecting the AE signal. In this example, gold leads in the shape of a bow-tie converged to the resistive element in the center of the hydrophone. The

maximum (90°) or minimum (270°) phase of the current injection waveform was synchronized with a square-wave pulse/receiver (Model 5077PR, Olympus NDT, Waltham, MA), which excited a single-element focused transducer (2.25 MHz, f-number 1.8, focal length 68.6 mm). Common mode noise was reduced by subtracting the two AE signals with opposite phases [Fig. 2(e)]. The AE signals, as well as simultaneously acquired pulse-echo signals (PE) received by the transducer, were amplified, band-pass filtered, and captured with a fast 12-bit acquisition board (Signatec Inc., Newport Beach, CA). A 2-D x - y raster scan provided a 3-D beam pattern based on the AE signals, which were further compared with simulations generated in Matlab (The MathWorks Inc., Natick, MA).

C. AE Hydrophone Simulation

1) Current Field Simulation—The electrical simulations were based on lead field theory. The current density distribution $\mathbf{J}(x, y, z)$ and electric fields were simulated using Matlab's partial differential equation toolbox. The injecting and detecting electrodes were defined as dirichlet boundary conditions, and the other boundaries had neumann boundary conditions. Finite element analysis was used to simulate the current density distribution with the same geometry as the experiments. The equation appropriate for a finite element simulation is the poisson equation $-\nabla \cdot (\sigma \nabla \phi) = \rho$ with the Neumann boundary condition $\mathbf{J} \cdot \mathbf{n} = 0$, where $\rho = \rho_0[\delta(\mathbf{r} - \mathbf{r}_1) - \delta(\mathbf{r} - \mathbf{r}_2)]$ [see Fig. 3(a)], is a dipole with point charges located at positions $\mathbf{r}_1$ and $\mathbf{r}_2$ within the sample or on the boundaries, to account for an injected current or detecting electrodes. An adaptive mesh refinement algorithm was also used to improve the accuracy of the current density distribution. As an example, Fig. 3(a) displays the simulated results for a bowtie hydrophone. After computation, the voltage and mesh information were exported from pdetool. The current density can be calculated by $\mathbf{J} = -\sigma \nabla V$, assuming $\mathbf{J} \cdot \mathbf{n} = 0$ at the edge of the hydrophone. In Fig. 3(b), the current density is highest at the center and near the electrodes.

2) Pressure Field Simulation—It is assumed that a single-element 2.25-MHz concave transducer (f-number 1.8, focal length 68.58 mm) is excited by the square pulse from a pulse/receiver, and the ultrasound pressure field $p(x, y, z, t)$ can be created in Field II simulation software [22], [23]. A complete map of an ultrasound beam pattern can, therefore, be obtained by scanning the ultrasound beam along the lateral and elevational directions over the surface of AE hydrophone. A 2-D simulation of the AE signal was created by convolving the ultrasound field $p(x, y, z, t)$ (simulated with Field II) with the inner product of the lead fields $\mathbf{J}^I$ and $\mathbf{J}^L$, which can be computed quickly using the 2-D fast Fourier transform algorithm.

3) AE Field Simulation—Combining the ultrasound beam and bowtie electric field resulted in the simulated AE image in Fig. 3(d). If the thickness of the AE hydro-phone is small compared with the ultrasound wavelength λ , the solution reduces to a double integral over the lateral and elevational directions at the depth of the hydrophone. If the AE hydrophone is oriented on the x - y surface plane, $w(x - x_0, y - y_0, z - z_0)$ becomes $w(x - x_0, y - y_0)$.

Dimensional analysis over thickness and width can be done along the x - z plane, such that $w(x - x_0, y - y_0, z - z_0)$ becomes $w(x - x_0, z - z_0)$. $\mathbf{J}^I$ is assumed constant and independent of position (x, z) over the cross-section, and its unit vector is the same as $\mathbf{J}^L$. Thus, $w(x, z) = K_I(x, z) \rho_0(x, z) |\mathbf{J}^I| |\mathbf{J}^L|^2$, because the connection between the electrode and the slab is usually the small area around the center of the cross-section. The detected V^{AE} will, therefore, be the average over the center area. In simulations, the electrodes are assumed to be point contacts.

D. Performance Analysis

A calibrated commercial hydrophone (HGL-0200, Onda Corp., Sunnyvale, CA) was used as a standard to calibrate the pressure from the ultrasound transducer and determine the pressure at the focus at different excitation voltages. The reference hydrophone was operated with the same hardware and software filtering characteristics as used during the AE hydrophone experiments to maintain consistency in the pressure estimates.

Each AE hydrophone was evaluated at four different pressure and bias voltage levels. At each pressure, the bias voltage through the AE hydrophone was determined by the applied voltage of the signal generator, variable from 20 V peak-peak (maximum) to 0 V (minimum or control). The peak-peak value of the filtered data was used to determine sensitivity, which was calculated by dividing the detected voltage by the pressure.

To compare AE hydrophones with different lateral dimensions, the sensitivity must be divided by the lateral area of the SZ, yielding units of nV/Pa/mm².

IV. Results

The effect of hydrophone shape, thickness, and lateral width on the SZ, along with the conductivity ratio, were first determined from simulations.

A. Simulation-Based Design Parameter Analysis

1) Effect of Hydrophone Shape—The shape of hydro-phone is critical for determining the current distribution, selectivity (or resolution), and sensitivity of the AE hydrophone for mapping the ultrasound beam pattern, as depicted in Fig. 4(b). The AE hydrophone with a rectangular shape has the smallest full-width at half-maximum (FWHM) and lowest side lobe along the lateral x -direction [Fig. 5(a)]. Along the y -direction [Fig. 5(b)], the difference between AE hydrophones and the modeled ultrasound beam is not obvious. The x - y cross-section at the focus is elliptical for the bowtie, but circular for the rectangular dumbbell design. Therefore, the bowtie ($\sigma_r = 1$) is not axially symmetric and severely distorts the beam pattern, but the rectangular dumbbell is axially symmetric and circular along the cross-section.

2) Effect of Thickness—The effect of thickness on sensitivity and spectrum are observed by changing the thickness H along the cross-section with constant width (depicted in Fig. 6(c), [26]). In the time domain, if the thickness is changed from $\lambda/25$ to $\lambda/2$, the amplitude increases linearly. When thickness is greater than $\lambda/2$, the two top peaks begin diverging, and the signal saturates [Fig. 6(a)]. The linear range of sensitivity versus thickness extends from 0 to one-half wavelength [Fig. 6(c)].

The thickness has an important effect on the spectrum of the AE signal: when it increases from 0, the first harmonic magnitude decreases and reaches a minimum at thickness $\lambda/2$. If thickness is greater than $\lambda/2$, the first harmonic begins increasing and shifting to a lower frequency. If thickness is larger than λ , the first harmonic component is lost, reducing the main lobe center frequency [Fig. 6(b)]. Therefore, the source of spectrum distortion is primarily due to the thickness of the SZ. The bandwidth of the AE hydrophone is determined by the thickness of the SZ. From Fig. 6(d), it is clear that the bandwidth decreases with increasing SZ thickness.

3) Effect of Lateral Width—In Fig. 7, the transducer center frequency was 2.25 MHz (wavelength 657 μm) and the slab width, W , was varied from 0.025 mm to 7.025 mm [25]. In the time domain, the peak positions of V^{AE} are kept constant [Fig. 7(a)] and the sensitivity reaches saturation when the lateral width is larger than the beam size [Fig. 7(c)]. For the frequency spectrum of the AE signal, the position of higher order harmonic components does not change with lateral size, and the magnitude decreases and converges to a steady value [Fig. 7(b)]. From Fig. 7(d), the bandwidth does not change with the width of SZ, or approximately 2.04 MHz for any width size for detecting the 2.25 MHz transducer. Therefore, only the thickness of SZ determines the bandwidth of the AE hydrophone.

4) Effect of Material Conductivity Ratio—The ideal x - y cross-sectional beam pattern at the center of the focal spot is displayed in Fig. 4(b1) at a dynamic range of 40 db. For the bowtie hydrophone, if $\sigma_r = 1$ [Fig. 4(a4)], the x - y cross section is elliptical [Fig. 4(b4)], but when $\sigma_r = 20$ [Fig. 4(a5)], the x - y cross section [Fig. 4(b5)] is circular and close to the ideal shape in Fig. 4(b1). Au/ITO ($\sigma_r = 20$) exhibits the best performance of all the ITO hydrophones [$\sigma_r = 1$, Figs. 4(b2)–4(b4)].

B. Pressure Field Reconstruction

Using the simulated ultrasound beam [created by Field II, Fig. 3(c) and Fig. 8(a)] and the hydrophone properties [Fig. 3(a)], the AE signal can be obtained from (4). The simulated AE signal (with $\sigma_r = 20$) in Fig. 8(b) is close to the ultrasound beam [Fig. 8(a)], if SZ is assumed to be a delta function at the focus. The pressure field can be reconstructed when the geometric properties and conductive ratio σ_r of the AE hydrophone are known [see (5)]. In Figs. 8(c)–8(g), the side lobes along y become obvious when $\sigma_r > 1$, and the second side lobe along x emerges when $\sigma_r = 10$. When $\sigma_r = 20$, the reconstructed beam pattern is almost the same as that of the ideal ultrasound field. This implies that if the geometric properties and conductivity ratio of the simulated hydrophone match well with the actual device, the reconstructed pressure is close to the actual ultrasound field. The pressure field is approximated using the phase and magnitude of the detected AE signal and (8).

C. Mapping the Ultrasound Beam

According to (8), the pressure field is proportional to the detected AE signal, so an AE hydrophone is able to map the ultrasound beam. In Figs. 9(a)–9(c), AE and PE signals of the bowtie Au/ITO AE hydrophone are co-registered in one volumetric image. The PE ultrasound image reveals the location of the current source at the center of the AE hydrophone.

The frequency bandwidth of the device determines the axial resolution. According to the simulations, it is mostly determined by the thickness; the lateral width of the SZ has only a minor effect. The optimal resolution of the transducer requires a hydrophone that is significantly thinner than the envelope of the waveform, or less than $\lambda/2$. In Figs. 10(a) and 10(b), the PE and AE responses are plotted along with the frequency responses for the 100% ITO rectangular dumbbell hydrophone ($\sigma_f = 1$, 75 μm square) having a thickness of 100 nm. A Blackman–Harris window is applied in the time domain to remove the ring artifact caused by the substrate. In Fig. 10(c), the frequency components of the AE signal are similar to those of the PE signal. The spectrum of the received PE signal includes both the transmit and receive transfer function of the transducer, which are assumed to be the same. Therefore, to compare spectra, the magnitude of the received PE signal must be divided by 2. In Fig. 10(d), the spectrum of the AE signal matches well the adjusted spectrum of the PE signal below $2.6f_0$, whereas for frequencies higher than $2.6f_0$, the adjusted magnitude is higher than the AE magnitude. Several reasons can explain this: 1) the reflection from the glass substrate of the AE hydrophone introduces harmonic components in the received PE signal; 2) the thickness of SZ will decrease the magnitude of high frequencies in the AE signal [Fig. 6(b)]; 3) the data acquisition card receiving the AE signal has limited bandwidth.

D. Sensitivity and SNR Analysis

When the ultrasound transducer was excited by a 400 V pulse, the commercial reference hydrophone detected a peak-peak signal of 15.03 mV. From the sensitivity chart provided by the vendor, the sensitivity of the reference hydrophone at 2.25 MHz was 30.4 nV/Pa. When scaled to area, this became 968 nV/Pa/mm². The peak pressure at the focus was, therefore, 494 kPa.

The sensitivity and SNR were determined for several hydrophones, although we focused the analysis on the dumbbell design because its performance was shown to be superior according to simulation and experiment. The transducer output pressures of 500, 375, 250, and 125 kPa were measured with the calibrated hydrophone. As depicted in Fig. 11(a), sensitivity of the AE hydrophone was linearly proportional to the injected bias current for each pressure level (slope = 0.079, 0.081, 0.084, and 0.096 nV/Pa/mA for 125, 250, 375, and 500 kPa).

Sensitivity is weakly related to the conductivity ratio. The sensitivity is a function of the size of the SZ and the magnitude of the bias current. The experimental data [Fig. 11(a)] indicates that the sensitivity was relatively flat across the range of ultrasound pressures that were tested. Because the conductivity varies linearly with externally applied pressure (see [17, Eq. (1)]), the sensitivity is weakly related to the conductivity of the SZ. Because the conductivity of the electrode is assumed constant, then the sensitivity is weakly related to the conductivity ratio between the SZ and electrode.

The sensitivity improvement caused by the detector shape is not as obvious as the conductivity ratio. At a transducer focal pressure of 500 kPa at 2.25 MHz, the 100% ITO dumbbell hydrophone ($\sigma_f = 1$, 75 μm square) generated a signal strength of 0.515 mV_{p-p} for a bias voltage of 20 V. This indicates a sensitivity of 1.04 nV/Pa [Fig. 11(a)]. Scaled for the

SZ area (75 μm square), however, the sensitivity is 185 nV/Pa/ mm^2 , about 20% of the 968 nV/Pa/ mm^2 of the reference commercial hydrophone. The sensitivity of the 100% ITO bowtie hydrophone ($\sigma_f = 1$, 200 μm square) was 0.977 nV/Pa. When the ITO electrode was replaced with gold, the detected AE signal was 0.753 mV_{p-p}, corresponding to a sensitivity of 1.52 nV/Pa. Thus, the conductivity ratio has a larger effect on sensitivity than the shape of the hydrophone.

SNR is defined as the ratio of powers between the acoustoelectric signal and background noise. The noise level was determined by disconnecting the pressure and bias current ($I = 0$, $p = 0$) across the detector and measuring the background power (square of the peak-to-peak voltage) during the same interval as when measuring the AE signal with an applied current and pressure ($|I| > 0$, $|p| > 0$). A Hamming filter centered at 2.25 MHz was applied to each single-shot measurement. The bandwidth of the filter was 2.48 MHz. The average SNR for 100 single-shot measurements was used to report SNR. According to Fig. 11(b), the SNR increased with applied current and pressure; at a focal pressure of 500 kPa and bias current greater than 6 mA, the slope of SNR to current was 0.559 mA^{-1} . This suggests that the SNR can be improved further by increasing the injected bias current.

E. Calibration of Conductivity Ratio

The conductivity ratio of Au/ITO can be calibrated using the simulated and experimental data for the bowtie hydrophone. The conductivity ratio does not significantly affect the center profile or beam width along the y -axis, so only the profile along the x -direction is affected by the conductivity ratio of Au/ITO. In Fig. 12(a), when σ_f is increased from 1 to 50, the side lobe decreases and converges to the ideal beam profile. At -6 dB [Fig. 12(b)], if the conductivity ratio of Au/ITO is higher than 20, the beam width approaches the experimental result (1.9 mm). This is consistent with the fact that the conductivity ratio of Au/ITO is over 20. The large variation of ITO conductivity comes from different preparations of ITO, which is a solid solution typically composed of 90% indium oxide (In_2O_3) and 10% tin oxide (SnO_2).

These results suggest that the accuracy of the AE hydrophone depends largely on the shape and conductivity ratio of the device, whereas its sensitivity is determined by the size of the SZ and amplitude of the injected bias current.

V. Discussion

Several properties of the AE hydrophone were examined using both simulation and experiment in an effort to optimize the design of the AE hydrophone. Sensitivity depended on the amplitude of the bias current, as well as the width and thickness of the SZ. Sensitivity was proportional to the thickness (within a half wavelength) and width (within a beam diameter) of the SZ.

Thickness of the active element affects overall resistance and bandwidth. In general, increasing thickness improves sensitivity, while reducing bandwidth. A thin device with small lateral dimensions will have a higher current density and, consequently, higher output voltage for a given resistance. Because the thickness of the ITO detectors in this study was

~75 nm and the acoustic wavelength used for testing was 0.67 mm, it is theoretically possible to dramatically improve the sensitivity without sacrificing bandwidth within the range of medical ultrasound frequencies (1 to 40 MHz). Given the measured sensitivity of the Au/ITO hydrophone of 1.52 nV/Pa, it is expected that, if the thickness of the AE hydrophone is more than 2 μm , its Sensitivity will exceed the performance of the commercial Onda hydrophone (30.4 nV/Pa) and still retain excellent bandwidth.

Although lateral resolution of the device is determined by the width of the SZ, axial resolution is determined by the length of the pulse wave packet (or envelope). For the transducer used in this study, the beam size is approximately 0.657 mm (axial) and 1.44 mm (lateral), which is close to the beam FWHM ≈ 1.9 mm measured by the AE bowtie hydrophone; an SZ thinner than 328 μm ($\lambda/2$) can capture its profile and maintain the bandwidth with minimal distortion. For a 40-MHz transducer, a hydrophone thickness of 20 μm would deliver a sufficient frequency response and still offer decent sensitivity.

There are several reasons that we focus on the conductivity ratio as opposed to the piezoresistive coefficient. For higher conductivity ratios, the current distribution around the sensitivity zone acts more like a delta function, and the detected AE signal is linearly proportional to the pressure field (7). The conductivity ratio affects the accuracy of the detector, which was observed in both simulation and experiment. The Au/ITO ($\sigma_r = 20$) hydrophone exhibits better performance than the 100% ITO hydrophones ($\sigma_r = 1$), and produces close to an ideal beam shape. The sensitivity is less affected by the conductive ratio, as the experimental data [Fig. 11(a)] indicated that the ultrasound pressure has a weak influence on the sensitivity, changing over a small range, and the conductivity variability depends on the applied pressure (see [17, Eq. (1)]).

The resistance of the sensitivity zone that is square with different lateral size ($W \times W$) is determined by the thickness (H), $R = \rho W/(W \times H) = \rho/H$. If thickness is increased, the resistance will decrease. Because current density (J) is assumed constant in the theory, the current ($I = J \times W \times H$) will increase and the applied voltage ($V = I \times R = J \times W \times \rho$) will be constant. Therefore, the average power consumption ($V^2/R = V^2 H/\rho$) will increase. In [26], we demonstrated that increasing the resistance had two effects: it decreased the current density at a fixed voltage and increased the voltage drop of the recorded AE signal. There was no net effect on the amplitude of the detected AE signal, but the device with higher resistance required less power.

The AE detector design also includes the electrodes by which the leads are attached. These electrodes and SZ layer can all be made with a single layer of ITO, reducing its width gradually or suddenly as it approaches the high current density of the SZ. Alternatively, gold electrodes could be applied to the ITO. Although this requires additional clean-room processing, the benefits are manifold: the sensitivity zone is more precisely defined; the resistance across the device is effectively that of the SZ because the gold-to-ITO conductivity ratio is over 20; it is easier to make the bowtie shape than to create a clean long thin gap for the dumbbell hydrophone, especially a tiny size of several micrometers designed for a high-frequency transducer; and lead wires are more easily soldered to gold

than to ITO. Therefore, there is tradeoff between the fabrication complexity and performance.

Table I compares the properties and performance of the AE dumbbell hydrophone with the commercial piezoelectric hydrophone (onda HGL-200). There are potential advantages of the microelectromechanical systems (MEMS)-based AE hydrophone, especially lower cost and potentially higher bandwidth resulting from the very thin substrate of the active element. Although the sensitivity of the AE hydrophone was considerably smaller at 2.25 MHz [185 versus 968 nV/(Pa·mm²)], we expect much higher sensitivity with an increased bias current and thicker active element closer to one-half the acoustic wavelength. The AE hydrophone also employed a mechanically stronger active element (ITO) than the device based on a piezoelectric polymer [polyvinylidene fluoride (PVDF)] according to their Young's moduli. Finally, the MEMS-based fabrication of the AE hydrophone facilitates fabrication of AE detectors with elaborate geometries for applications in imaging and therapy.

In addition to characterizing the acoustic field of transducers, AE hydrophones also have potential for monitoring ultrasound exposure during therapy, such as high-intensity focused ultrasound (HIFU), lithotripsy, and hyperthermia, for which undesirable bioeffects are a serious concern, and the acoustic intensity is sufficient to destroy expensive detectors. AE ultrasound detectors could also operate as ultrasound receivers for photoacoustic imaging [27], where the acoustic signals are generated by the absorption of a high-energy laser pulse. The potentially low cost, high bandwidth, and robustness of AE hydrophones make them particularly attractive for many biomedical applications.

VI. Conclusion

This study examined design properties of the AE detector in an effort to optimize bandwidth and sensitivity for mapping an ultrasound beam pattern. These properties included the material composition, geometry (thickness and width) of the active element, and electrical characteristics (conductivity ratio). The AE hydrophone with a rectangular dumbbell configuration provided the best images of the ultrasound transducer beam pattern compared with other geometries. A conductivity ratio of 20 or higher produced the strongest AE signals, suggesting that ITO (active element) combined with gold (passive electrodes) would perform best as AE detectors. The sensitivity of a device fabricated with this design was determined to be 1.52 nV/Pa at 2.25 MHz, which would likely improve using a larger bias current or thicker active element. In the future, other properties, such as the acceptance angle and limits of the spectral response, will be investigated. The MEMS fabrication also enables new possibilities for AE detector arrays. Low-cost AE detectors may be valuable for characterizing the beam pattern of medical ultrasound transducers designed for imaging or therapy with advantages over more common piezoelectric devices.

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Biography

Zhaohui Wang (M'10) received his B.S. degree in electronics in 1993 from Shandong University, China; his M.E. degree in circuits and systems in 2002 from University of Science and Technology of China, China; his M.S. degree in biomedical engineering in 2005 from University of Toledo; his M.S. degree in mechanical engineering in 2008 from University of Arizona; and his Ph.D. degree in electrical and computer engineering in 2011 from the University of Arizona. He is now a postdoctoral Fellow at the University of Pittsburgh, Pittsburgh, PA. His current research includes the acoustoelectric effect and photoacoustic imaging.

Pier Ingram received his B.A. degree from the University of Oxford and his M.S. degree in optical sciences from the University of Arizona. As a research specialist, he is able to exercise a variety of skills including electronics, coding, testing, and experimentation. He has been with the Experimental Ultrasound and Neural Imaging Laboratory since its inception in 2007.

Charles Greenlee was born in 1982 in the industrious city of Gary, IN. In 2005, he received a B.S. degree with a double major in optical engineering and electrical engineering from Rose-Hulman Institute of Technology in Terre Haute, IN. As an undergraduate, he participated in numerous internships which helped nurture his passion for product design and research and development. Upon completion of his undergraduate studies, he began

graduate school at the University of Arizona, where he was awarded an M.S. degree in optical sciences in 2012. His research interests include optical thin-film interference filter design and characterization, cleanroom microfabrication, and characterization and incorporation of electro-optic polymers in next-generation free-space photonic devices. He is currently employed as a Senior System Application Engineer at Nanometrics Inc. in Milpitas, CA.

Ragnar Olafsson received the B.S. degree in electrical engineering in 2002 from the University of Iceland, Reykjavik, Iceland. In 2002, he received a Fulbright scholarship to pursue a graduate degree in biomedical engineering at the University of Michigan, Ann Arbor, MI, where he completed the M.S. and Ph.D. degrees in 2004 and 2008, respectively. From 2008 to 2010, he worked as a postdoctoral researcher with the Department of Radiology in the College of Medicine at the University of Arizona, Tucson, AZ. He is currently Senior Electrical Engineer at Acutus Medical in San Diego.

Robert A. Norwood received the B.S. degree in physics and mathematics from the Massachusetts Institute of Technology in 1983 and the Ph.D. degree in physics from the University of Pennsylvania in 1988. He subsequently led R&D groups at Hoechst Celanese and Alliedsignal (Honeywell); his group at Alliedsignal developed AErospace-qualified polymer waveguide technology that was the best in the world at the time, and he helped to secure the sale of this business to Corning Photonics in 1999. Dr. Norwood was Vice-President and Chief Technology Officer at Photon-X Inc., a venture-capital-funded photonics startup company based in Malvern, PA, and started in 1999; the company set the record for the lowest-loss single-mode polymer waveguides ever developed at 1550 nm. Dr. Norwood is now a Professor in the College of Optical Sciences at the University of Arizona, where he performs research on high-speed electro-optic modulators and switches, integrated magneto-optic devices, polymer-based integrated optics, 3-D display technology, nanoimprinting, organic photovoltaics, nonlinear optical fibers, optical microresonators and ultrafast optical switching, among other areas. He is a world expert in polymer-integrated optics and optical materials, with more than 85 refereed publications, 6 book chapters, 29 issued US patents, and 50 invited talks. Dr. Norwood has served on the program committee

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Russell Witte received the B.S. degree in physics with honors from the University of Arizona, Tucson, AZ, in 1993 and the Ph.D. degree in bio-engineering from Arizona State University, Tempe, AZ, in 2002. His graduate work exploited chronic, implantable microelectrode arrays to characterize sensory coding, learning, and cortical plasticity in the brain. He then moved to Ann Arbor and the University of Michigan, where he developed noninvasive imaging techniques to explore the neuromuscular and nervous systems. While at the Biomedical Ultrasound Laboratory, he helped devise several methods for enhancing image contrast using ultrasound combined with other forms of energy. He then made a bold move in 2007 and accepted a position as Assistant Professor of Radiology, Biomedical Engineering and Optical Sciences at the University of Arizona. His laboratory currently develops novel techniques that integrate light, sound, and electricity for medical imaging. His research potentially impacts applications ranging from epilepsy and arrhythmia to cancer.

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Fig. 1.

(a) Top and side view of the bowtie Au/ITO AE hydrophone built on a glass substrate. The conductive area is made of gold, and the material in the sensitivity zone is ITO. The current is injected from a current source electrode to a current sink electrode, and the AE signal is obtained from the electrodes V+ and V-. (b) Schematic of the AE effect on the bowtie hydrophone design. $H(x_0, y_0, z_0)$ is at the center of the sensitivity zone and the center of the transducer is $C(0, 0, 0)$, and any point P in the ultrasound pressure field (x, y, z) can be described in the electric field as $(x - x_0, y - y_0, z - z_0)$, or $\vec{HP} = \vec{CP} - \vec{CH}$.

Fig. 2.

Experimental setup for characterizing the hydrophones. A current source injected a 200 Hz rectangular waveform into the hydrophone. AE signals were acquired at the maximum (90°) and minimum (270°) phase of the waveform. Top right: Pulse-echo (PE) signal. Bottom right: The AE signals acquired at the different phases of the injected current waveform. Common mode noise was reduced by subtracting the two AE signals with opposite phase. The US transducer has $f_0 = 2.25$ MHz, $f = 68.58$ mm, f-number = 1.8.

Fig. 3.

Acoustoelectric (AE) signal induced by the 2.25-MHz transducer for a bowtie hydrophone design with a center gap of $200\ \mu\text{m}$. The parameters of simulation were consistent with the experimental setup. (a) Current distribution and electric field were simulated by conductive media dc application in Matlab's partial differential equation toolbox, where two electrodes were defined by Dirichlet boundary conditions, the other boundaries by Neumann boundary conditions. (b) The current density distribution of lead field was calculated based on the simulated potential from the electric field and mesh information. (c) The acoustic field was simulated using Field II software. (d) Simulated AE envelope on a decibel scale at the focus of the bowtie hydrophone with center dimensions of $200 \times 200\ \mu\text{m}$. The images were plotted with range $[-40, 0]\ \text{dB}$ along the x - y plane in millimeters.

Fig. 4.

Simulation results for hydrophones with different shapes and conductivity ratios. The acoustoelectric (AE) signals induced by a 2.25-MHz transducer are simulated with four different hydrophone designs with center area of 200 μm square (top row). The bottom row contains simulated images of the AE signal along the x - y plane at the focus for the actual beam pattern, circular dumbbell ($\sigma_r = 1$; a2), rectangular dumbbell ($\sigma_r = 1$; a3), bowtie ($\sigma_r = 1$; a4) and bowtie ($\sigma_r = 20$; a5). The decibel range is 40 dB, and the horizontal and vertical crossing lines mark the position of three coordinates. (b4) has the highest side lobe and largest beam width along the x -direction. Field II was used for all ultrasound simulations.

Fig. 5.

(a) Acoustoelectric (AE) envelope profile along the x -direction at the focus; (b) Profile along the y -direction. The AE hydrophone with the rectangular shape has the smallest FWHm and smallest side lobe along the x -direction. “Beam” denotes the simulated transducer beam pattern from the detected pulse–echo (PE) signal.

Fig. 6.

Simulation: Effect of thickness (H) of the sensitivity zone, with H varied from $\lambda/25$ to λ . (a) Simulated acoustoelectric (AE) signals at the center of the cross-section. (b) Power spectrum of the AE signals with range -70 dB. (c) Plot of sensitivity approaching a saturation value that is dependent on wavelength, where $\lambda/10$ is used as the evaluation unit of thickness. The center frequency of the transducer is 2.25 MHz (wavelength $\lambda = 657 \mu\text{m}$), and the width (W) of the sensitivity zone (SZ) is 0.114λ . (d) Bandwidth of (b) versus thickness of SZ.

Fig. 7.

Simulation: Effect of lateral width (W) of the sensitivity zone, with W varied from $\lambda/25$ to 6λ [same device as that shown in Fig. 6(c)]. (a) Simulated acoustoelectric (AE) signals and (b) AE power spectrum with a range of -70 dB. (c) Plot of sensitivity with the width of the sensitivity zone (SZ), and the saturation value is limited by the beam size with a full-width at half-maximum (FWHM) of 1.8λ . (d) Bandwidth (FWHM) of (b) versus width of SZ.

Fig. 8.

The reconstructed ultrasound beam compared with the ideal ultrasound field using different values of the conductivity ratio ($\sigma_r = 1, 2, 5, 10, 20$). Comparison and analysis are made along the x (row 3) and y profiles (row 4) at the center of the beam. Column (a) is the simulated ultrasound field, column (b) is the simulated bowtie AE hydrophone with $\sigma_r = 20$, columns (c)–(f) are the reconstructed ultrasound pressure field from the simulated Au/ITo hydrophone AE signals [column (b)]. Dynamic range for the images is 40 dB and the scale for plots is 0 to –70 dB. Full-width half-maximums (–6 dB) are displayed with a red arrow on the profile plots.

Fig. 9.

Experimental results: The ultrasound beam of the 2.25-MHz ultrasound transducer was mapped using a bowtie Au/ITO [sensitivity zone (SZ) = 1 mm] acoustoelectric (AE) hydrophone (d) at the focus (68 mm). (a)–(c) The 3-D cutaway view of AE ([−12, 0] dB, hot colors, gold + ITO) and pulse–echo (PE) ([−22, 0] dB, gray, the glass substrate) images along x - y and x - z planes at the center of the AE hydrophone. The blue cross hairs depict the pixel in common for all three planes (a)–(c). The AE and PE signals are co-registered automatically because of the one-way versus two-way travel of the acoustic wave. (e) 3-D rendering of the AE image ([−22, 0] dB) describes the ultrasound beam at the focus. Side lobes are clearly visible around the focal spot, but slightly delayed in the depth direction.

Fig. 10.

Experimental results: Spectrum analysis of dumbbell ($\sigma_r = 1$, $SZ = 75 \mu\text{m}$ square) acoustoelectric (AE) hydrophone. (a) and (b) are the time sequence and spectrum of the pulse-echo (PE) signal and the corresponding AE signal obtained at the center of the AE hydrophone. (c) To reduce the ring effect from the substrate, both AE and PE signals are processed with a Blackman-Harris window in the time domain. (d) The transmit and receive transfer functions of the ultrasound transducer are assumed to be the same. To compare the spectra between the filtered PE and AE signals, the magnitude of the received PE signal was divided by 2.

Fig. 11.

Experimental results: Sensitivity (a) and SNR (b) analysis of rectangular dumbbell ($\sigma_r = 1$) hydrophone show the effect of injected current and transducer pressure. The sensitive zone of the dumbbell ($\sigma_r = 1$) is $75 \mu\text{m}$ square. The center frequency of the transducer is 2.25 MHz. The resistance between two injecting electrodes is $1.8 \text{ k}\Omega$. The slope of sensitivity versus current at pressures of 125, 250, 375, and 500 kPa are 0.079, 0.081, 0.084, and 0.096 nV/Pa/mA, respectively.

Fig. 12.

The effect of the conductivity ratio of Au/ITo on the acoustoelectric (AE) bowtie hydrophone beam pattern. “Experimental” is from the experimental data of bowtie hydrophone with ITO on the 1×1 mm center and gold on the remainder. “Beam” is the transducer beam pattern at the focus simulated by FIELD II, “Simul. σ_r ” is the simulated AE signal at the focus with different σ_r of Au/ ITO. (a) Profile on the center along the x -direction, where experimental has -30 dB noise floor caused by noise; $\sigma_r = 1$ has the highest side lobe and largest full-width at half-maximum (FWHM), but when σ_r increases, the AE profile converges to the transducer beam pattern. (b) Conductivity ratio of Au/ITo versus beam width. “ $x-\sigma_r$ ” is the beam width at -6 dB along the x -direction for simulated hydrophones with different conductivity ratios. “ x -experiment” is the beam width at -6 dB along the x -direction for real Au/ITo hydrophone. “ x -beam” is the beam width at -6 dB along the x -direction for the simulated transducer beam pattern, created by the convolution of ultrasound field with the pulse–echo (PE) signals. At -6 dB, if the conductivity ratio of Au/ITo is higher than 20, the beam width converges and is close to the experimental result (1.9 mm, “ x -experiment”).

TABLE I

Comparison Between Acoustoelectric (AE) and Piezoelectric Hydrophones.

Parameters	Onda HGL-200	AE dumbbell
Mechanism of action	Piezoelectric	Acoustoelectric
Active material	PVDF [28]	ITO
Young's modulus of active material	~1 GPa [29]	180 GPa [30]
Active area (mm ²)	0.04	0.0056
Thickness of active region (μm)	~30	~0.1
Sensitivity (at 2 MHz) (nV/(Pa-mm ²))	968	185
Bandwidth (MHz)	40 [28]	>1000
Approximate cost	\$3000	<\$100