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Inexpensive Acoustoelectric Hydrophone For Mapping High Intensity Ultrasonic Fields

Russell S. Witte, Ph.D.,

Department of Radiology, University of Arizona, Tucson, AZ 85724

Tim Hall, Ph.D.,

Department of Urology Surgery, University of Michigan Health System, Ann Arbor, MI 48109

Ragner Olafsson, Ph.D.,

Department of Biomedical Engineering, University of Michigan, Ann Arbor, MI 48109

Sheng-Wen Huang, Ph.D., and

Department of Biomedical Engineering, University of Michigan, Ann Arbor, MI 48109

Matthew O'Donnell, Ph.D.

Department of Bioengineering, University of Washington, Seattle, WA 98195

Abstract

We describe an inexpensive alternative to conventional hydrophones for measuring ultrasonic fields. The hydrophone, composed of common laboratory supplies, depends on the acoustoelectric (AE) effect, a well-known interaction between electrical current and pressure. Beam patterns of a 540 kHz annular transducer captured using a bowtie graphite hydrophone were consistent with patterns obtained using conventional, more expensive hydrophones. The AE signal was proportional to both the applied bias current (1.83 $\mu\text{V}/\text{mA}$) and pressure (13.3 $\mu\text{V}/\text{MPa}$) with sensitivity better than 50 kPa. Disposable AE hydrophones may be an attractive alternative for clinical applications that require close monitoring of high intensity acoustic fields.

I. INTRODUCTION

As clinical applications for ultrasound therapy continue to proliferate--from lithotripsy to ablation treatment of uterine fibroids and cancer--the need for simple, rapid, and accurate estimates of acoustic fields become increasingly important. Although conventional acoustic measuring devices, such as the membrane, needle and fiber optic hydrophones, have been successfully employed in a variety of medical procedures, each has drawbacks¹⁻³. They exhibit low sensitivity, reduced bandwidth, high damage susceptibility, and/or cost several thousand dollars or more. As an alternative to traditional hydrophones, we describe early prototypes of a device that maps an ultrasound beam pattern based on the acoustoelectric (AE) effect. The device has attractive attributes not typically seen with other hydrophones: simple construction, low cost (disposable), decent sensitivity, and potential resistance to damage at high intensity ultrasonic fields. Initial findings were reported in a recent conference paper.⁴

II. BACKGROUND AND THEORY

The AE effect is based on an interaction between a propagating acoustic wave and charged particles in a material, such that

$$\frac{\Delta\rho}{\rho_0} = -k\Delta P, \quad (1)$$

with material electric resistivity ρ , incident pressure wave P , and interaction constant k . Previous hydrophones based on the AE effect have relied on exclusively piezoelectric materials embedded between two semiconductors to extract measures of ultrasound propagation.^{5–7} Those devices were capable of only measuring acoustic power without phase information and had low sensitivity—45 dB less than that of efficient piezoelectric transducers.⁸ We describe a new type of AE hydrophone based on a bias current i injected across a conducting material. As ultrasound propagates through a uniform conducting element, a voltage signal V^{AE} signal can be detected between two recording electrodes according to Ohm's law and the AE effect. Biomedical applications based on this concept range from electrical impedance imaging⁸ to mapping biological current flow.^{9–11} The AE hydrophone exploits a small region of high current density J (the “sensitivity zone”) separated by a resistivity ρ_0 to map an ultrasound beam b . The following 4D expression, based on a recent derivation of the general AE equation, assumes far field detection of the AE signal:

$$V^{\text{AE}}(x,y,z,t) = -k\rho_0 P_0 \iiint J(x,y,z)b(x,y,z)a\left(t - \frac{z}{c}\right)dx dy dz, \quad (2)$$

with pressure pulse amplitude P_0 , pulse waveform $a(t)$, and speed of sound c .^{12,13} As the sensitivity zone becomes thin compared to the ultrasound wavelength λ , the solution reduces to a double integral over the lateral extent of the ultrasound pressure wave at the depth of the sensing element with V^{AE} sensitive to BOTH the phase and magnitude of the incident acoustic field. A complete map of an ultrasound beam pattern can, therefore, be obtained by scanning the ultrasound beam relative to the hydrophone.

We designed, constructed and tested a variety of inexpensive AE hydrophones composed of common laboratory supplies. Simple hydrophone geometries and the basic configuration for producing and measuring the AE signal are displayed in Fig. 1. An alternating bias current (~ 100 Hz) was applied across the sensitivity zone between two AgCl electrodes; the oscillating field limited adverse effects associated with direct current, such as electrode polarization and hydrolysis. An ultrasound transducer was pulsed at the peak or trough of the bias current, and the voltage signal was recorded on a second pair of electrodes, differentially amplified, high-pass filtered to isolate the ultrasound modulated component, and captured on a 12-bit 100 MHz acquisition board (Signatec Inc.). The AE signals generated at $+i$ and $-i$ were subtracted to reduce common mode interference. Feedback on the position and geometry of the hydrophone were obtained from co-registered pulse echo signals. A calibrated custom-built fiber-optic hydrophone¹⁴ and commercial needle hydrophone (Precision Acoustics Inc.) provided reference maps of the ultrasound beam pattern for direct comparison to the AE devices. The ultrasound source was a 540 kHz single-element annular transducer driven by a pulse-receiver (5077PR, Panametrics Inc.); a raster scan of the ultrasound transducer produced a focal beam pattern for each hydrophone.

III. RESULTS

An initial bowtie hydrophone was hand-drawn with graphite on standard inkjet paper. A conductive gel (2% Agarose, 2% NaCl in diH₂O) was added on top of the graphite away from the sensitivity zone to improve electrical coupling between the electrodes and graphite; mineral oil covered the top layer of the hydrophone for electrical isolation and enabled ultrasound to pass through the conductive layer with minimal surface reflections. Fig. 2 depicts sample A-lines (A, B) and power spectra (C) of the pulse echo and acoustoelectric signals. The thickness of the sensitivity zone was small compared to the ultrasound wavelength and faithfully reproduced the frequency response of the 540 kHz transducer (C). As predicted by Eq. 2, the AE signal was linearly proportional to both the bias current and acoustic pressure over a wide operating range; Fig 2D illustrates the amplitude of the AE signal at five ultrasound pressures and ten bias current levels. The slopes of the best-fit line of the AE signal were 13.3 $\mu\text{V}/\text{MPa}$

at 19 mA and $1.83 \mu\text{V}/\text{mA}$ ($r^2=0.99$) at 2 MPa. The noise equivalent pressure was approximately 40 kPa at 19 mA. A scan of the transducer also produced beam profiles of the ultrasound transducer. The beam profile for the AE device is similar to images obtained using more expensive hydrophones (see Fig. 3 for a direct comparison) with excellent sensitivity. The measured axial and lateral resolutions (5.2 and 4.1 mm, -6 dB) were within $400 \mu\text{m}$ for each of these hydrophones. In addition, the lateral beam profiles at the ultrasound focus exhibited side lobes (see Fig. 4 and Multimedia 1) consistent with the geometry of the ultrasound transducer.

A second hydrophone was constructed from standard off-the-shelf products. Two square pieces of copper tape ($50 \mu\text{m}$ thickness) were separated by a Panasonic thick film resistor (#0402) forming the shape of a dumbbell and placed on a thin Mylar® substrate. The device was coated with a water-resistant, insulating thin layer of rubber cement, enabling the device to be completely submerged in water for testing. A single pair of electrodes was used to pass current and record the AE signal. Figure 5 depicts beam patterns near the ultrasound focus using two different resistors ($0.2 \text{ k}\Omega$ and $10 \text{ k}\Omega$) with the same geometry. Although the larger resistor produced a greater voltage drop between the electrodes, the system was limited by the voltage source, producing a proportional drop in bias current. The net effect was no appreciable signal gain or loss as a result of swapping resistors. The $10 \text{ k}\Omega$ hydrophone was effectively a lower power device, which may be especially important for *in vivo* applications in which large current sources are undesirable. The axial and lateral beam sizes were $5.7 \text{ mm} \times 4.6 \text{ mm}$, slightly larger than the images produced by the bowtie-graphite hydrophone. The difference is likely due to the size and thickness of the resistor ($1 \times 0.5 \text{ mm}$, 0.35 mm) and relatively larger dimensions of the sensitivity zone.

IV. CONCLUSION

We have described a new type of disposable hydrophone that couples acoustic pressure into electrical energy based on the AE effect. The device relies on purely conductive, rather than piezo, materials to detect ultrasound. Its relatively simple construction from common laboratory supplies is especially appealing for a variety of medical applications that require close monitoring of ultrasonic exposure. The measured signals of these early prototypes were proportional to both the acoustic pressure and bias current, faithfully reproducing the ultrasonic beam pattern and spectrum of a low frequency transducer with a sensitivity less than 50 kPa. Future prototypes will push the limits of sensitivity and bandwidth and determine whether high frequency harmonics can be detected during ultrasound exposure. The inexpensive AE hydrophone may be an attractive alternative to commercial devices designed to map high intensity acoustic fields for a variety of medical applications.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

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Fig. 1.

Sample hydrophone designs and apparatus for mapping acoustic field patterns. Electrodes ($S_{\pm}$, $R_{\pm}$) were used to apply a bias current (I) and detect the AE voltage signal (V). Each hydrophone featured a small sensitivity zone capable of high current density that was used to selectively map the ultrasound beam based on the AE effect. For this study, a 540 kHz single element ultrasound transducer (Etalon Inc., focal length=90 mm, $f/\# = 1$) was used as an acoustic source. The right column depicts prototype bowtie and dumbbell designs used in this study, as well as a photograph of an actual device in the apparatus [FPGA = field programmable gate array; DAQ = digital acquisition; Trig = Trigger, S=Stimulation, R=Recording].

Fig. 2.

(A) Pulse echo (PE) reflections from the hydrophone and oil/air interface near the ultrasound focus. (B) Simultaneously acquired acoustoelectric (AE) signals at the same position. The solid red line indicates a positive current injection, whereas the dashed black line denotes a negative current injection and consequent 180° phase shift of the AE signal. (C) Comparison of the frequency spectrum between the AE (dashed red) and PE (solid black) signals. Both are centered near 500 kHz with similar bandwidth. (D) Linear effect of pressure and current on the AE signal. At maximum pressure (2 MPa) the AE signal increased monotonically with current ($1.83 \mu\text{V}/\text{mA}$, $r^2=0.99$), whereas at maximum current (15 mA) the AE signal increased linearly with pressure ($13.3 \mu\text{V}/\text{MPa}$, $r^2=0.99$).

Fig. 3.

Comparison of axial ultrasound beam profiles (XZ) using bowtie-graphite AE hydrophone (left), needle hydrophone (middle) and fiber optic hydrophone (right). (A) Filtered RF signal near the ultrasound focus; (B) Magnitude of the complex signal on a log scale (6 dB dynamic range). The full width half maximum (FWHM) of 5.2 mm (axial) and 4.1 mm (lateral) determined by the AE device were within 400 μm of either of the other two more expensive hydrophones.

Fig. 4. Lateral beam profiles (XY) from the bowtie-graphite and needle hydrophones depicting side lobes. (left) Image of RF signal at a depth near focus; (right) Magnitude of line through center of lateral profile. Also see Multimedia 1 for a 13- μ sec movie of the XY profile as ultrasound beam passes through hydrophone.

Fig. 5. Ultrasound beam profiles (XZ) from a submersible dumbbell-copper, thick film resistor hydrophone.