Ultraminiature AlN diaphragm acoustic transducer

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Alison E. Hake,¹ Chuming Zhao,¹ Lichuan Ping,² and Karl Grosh^{1,3,a)}

AFFILIATIONS

¹Mechanical Engineering Department, University of Michigan, Ann Arbor, Michigan 48109, USA

²Singular Medical USA, Irvine, California 92614, USA

³Biomedical Engineering Department, University of Michigan, Ann Arbor, Michigan 48109, USA

^{a)}Author to whom correspondence should be addressed: grosh@umich.edu

ABSTRACT

Piezoelectric acoustic transducers consisting of a circular aluminum nitride and silicon nitride unimorph diaphragm and an encapsulated air-filled back cavity are reported. Analytical and finite element analysis models are used to design the transducer to achieve low minimum detectable pressure (MDP) within chosen size restrictions. A series of transducers with varying radii are fabricated using microelectromechanical systems (MEMS) techniques. Experimental results are reported for a transducer with a $175\text{ }\mu\text{m}$ radius on a $400 \times 500 \times 500\text{ }\mu\text{m}^3$ die exhibiting structural resonances at 552 kHz in air and 133 kHz in water. The low-frequency (10 Hz–50 kHz) sensitivity is $1.87\text{ }\mu\text{V}/\text{Pa}$ ($-114.5\text{ dB re } 1\text{ V}/\text{Pa}$) in both air and water. The sensor has an MDP of $43.7\text{ mPa}/\sqrt{\text{Hz}}$ (67 dB SPL) at 100 Hz and $10.9\text{ mPa}/\sqrt{\text{Hz}}$ (55 dB SPL) at 1 kHz. This work contributes a set of design rules for MEMS piezoelectric diaphragm transducers that focuses on decreasing the MDP of the sensor through size, material properties, and residual stress considerations.

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Developments in microfabrication technologies have enabled the miniaturization of pressure and acoustic transducers, leading to successes in consumer electronics applications through increased yield and decreased cost.^{1–4} These successes have also facilitated the design of implantable sensors for biomedical applications including intracochlear, intraocular, bladder, intra-abdominal, and cardiovascular pressure sensing.^{5–7} Two key transducer specifications required for inner ear acoustic pressure sensing include low minimum detectable pressure (MDP) and small sensor size. Achieving these specifications may result in a sensor that can viably be incorporated in a completely implantable cochlear implant system.

Capacitive, piezoresistive, and piezoelectric transduction, which can all be realized with microelectromechanical systems (MEMS) fabrication technologies, have been employed in attempts to meet the requirements for these and other applications. Piezoelectric sensors, in particular, have some notable benefits. First, they offer high sensitivity and respond linearly over a wide dynamic range.¹ Furthermore, they can achieve low power consumption because they do not require direct current (DC) bias voltage during operation^{1,8} and they can be rapidly power cycled because a charge pump is not needed for operation. Another key benefit for biomedical applications is that they can be made of biocompatible materials.⁹ The goal of our approach is to design a piezoelectric diaphragm sensor with a reduced diaphragm area, chip size, and a smaller MDP than current sensors, yet maintains a large dynamic range.

To begin, we consider an acoustic transducer consisting of a deformable circular diaphragm with an air-filled encapsulated back cavity, illustrated in Fig. 1(a). The device cross section in Fig. 1(b) shows the unimorph diaphragm with an aluminum nitride (AlN) piezoelectric layer and a silicon nitride (Si_3N_4) structural layer. AlN is the chosen piezoelectric material due to its high electrical resistance, low dielectric loss, and biocompatibility for *in vivo* use.^{9–12} The AlN and Si_3N_4 thicknesses, t_{AlN} and $t_{\text{Si}_3\text{N}_4}$, are designed to set the neutral axis of the layered diaphragm close to the AlN/ Si_3N_4 interface to maximize the voltage output for this configuration. This ensures that there is no charge cancellation within the AlN layer due to opposite strain directions above and below the neutral axis while keeping the thickness as small as possible increases diaphragm compliance and sensitivity.¹³ The cross-sectional view also demonstrates the sensor electrode configuration. In the most general configuration, there is a center circular electrode with radius r_1 and a ring electrode with inner radius r_2 and outer radius r_t that coincides with the total diaphragm radius. The low-frequency deflection of a clamped circular plate has an opposite curvature over the two electrode regions; therefore, flipped electrical polarity is used for a summed output. Eliminating (or disconnecting) either the center or ring electrode leaves a single-electrode design. In this paper, we examine a single outer ring electrode design for simplified fabrication and testing.

Analytical modeling of the configuration is used to design for the smallest possible MDP, calculated as the quotient of the square root of

FIG. 1. (a) A 3D rendering showing the transducer chip ($700 \times 600 \times 500 \mu\text{m}^3$), indicating the diaphragm with electrodes and an encapsulated back cavity. (b) The diaphragm cross section illustrating the AlN ($t_{\text{AlN}} = 1.0 \mu\text{m}$) and Si_3N_4 ($t_{\text{Si}_3\text{N}_4} = 1.1 \mu\text{m}$) structural layers and electrode configuration. (c) The fabricated transducer with $r_t = 175 \mu\text{m}$ and $r_2 = 145 \mu\text{m}$ (active outer electrode). The inner electrode is deposited but not connected to the output (inactive), and the total MEMS die size is $400 \times 500 \times 500 \mu\text{m}^3$. (d) The transducer after wire-bonding and silicone encapsulation.

output noise spectral density and the transducer sensitivity. The residual stress in the diaphragm is neglected in the initial design stage. Additionally, we disregard the diaphragm interaction with the back cavity volume because the acoustic compliance of the back cavity is more than 100 times larger than that of the diaphragm structure for the chosen geometry. The governing equation for the symmetric bending of a clamped circular plate due to uniform pressure loading is solved to obtain the equation for the diaphragm deflection.¹⁴ The deflection profile determines the strain in the diaphragm, which is used in the strain-charge form of the piezoelectric coupling equations to determine the open circuit voltage sensitivity to incoming pressure of the sensor.¹⁰

The intrinsic noise spectral density due to the dielectric losses of the piezoelectric material is considered to be the primary noise contribution and the lower limit for the system. It is calculated as $4k_b T \tan \delta / (\omega C)$, where k_b is the Boltzmann constant, $\tan \delta$ is the loss tangent of the material, ω is the frequency, and C is the device capacitance. For example, the noise at 1 kHz of a 3 pF AlN sensor with a loss tangent of 0.2% is $42 \text{ nV}/\sqrt{\text{Hz}}$, which is higher than the input referred noise of typical field effect transistors and operational amplifiers used to amplify the signal. Other sources of noise, such as electrical circuit noise, will add to this limit and increase the output noise. The resulting MDP at low frequencies for a one-electrode outer ring design with units of $\text{Pa}/\sqrt{\text{Hz}}$ is

$$\text{MDP} = 32D\tilde{k}\sqrt{\frac{k_b T \tan \delta}{\epsilon_{33}\pi\omega t_{\text{AlN}}^3}} \frac{1}{r_2^2 \sqrt{r_t^2 - r_2^2}}, \quad (1)$$

where $\tilde{k} = (\epsilon_{33}s_{11}^E(1 - \nu) - 2d_{31}^2)/d_{31}$ (ν is the Poisson ratio, ϵ_{33} is the dielectric constant, s_{11}^E is the compliance, and d_{31} is the piezoelectric coefficient) and D is the flexural rigidity of the layered diaphragm.¹⁰ In this equation, the chosen piezoelectric material and layer thickness dictate all parameters aside from ω , r_b , and r_2 . Equation (1) can be used to show that an r_2 to r_t ratio of 82% minimizes the MDP and, at the optimal radii ratio, the MDP is proportional to $A_d^{-1.5}$ as r_t is increased.

Most fabricated devices have residual stress, and so we also assessed the impact of this effect. The addition of residual stress modifies the governing equations for the system as given in Timoshenko and Woinowsky-Krieger.¹⁴ For instance, if there is 50 MPa of tensile stress in the piezoelectric layer of the diaphragm (possible with different types of film deposition^{12,15}) the MDP will be 2.5 dB higher than the no-stress case (using 18% electrode coverage). The optimal

electrode coverage for 50 MPa residual stress is 16%, which only provides about 1% MDP improvement over 18% coverage. Therefore, we continued to use the no-stress optimum electrode coverage for the sensor design.

To more fully understand the fluid-structure interaction and frequency behavior of the optimized device in air and water, a 2D axisymmetric finite element analysis (FEA) model is created with COMSOL Multiphysics. The structural and fluid material properties are chosen from the COMSOL material library,¹⁶ except the piezoelectric coefficient where we used $d_{31} = -2.51 \text{ pm/V}$, a value present in high quality AlN films.¹⁷ We simulated the transducer response to sound holding $t_{\text{AlN}} = 1 \mu\text{m}$ and $t_{\text{Si}_3\text{N}_4} = 1.1 \mu\text{m}$ constant and varying r_t between 100 and $275 \mu\text{m}$. The resulting resonant frequency, sensitivity, and MDP for each radius are tabulated in the [supplementary material](#). The numerical calculations confirm the predicted MDP/diaphragm area relationship given by Eq. (1) and indicate the expected decreased resonant frequency due to water loading. We characterize the sensor in water since it will be used in a submerged environment, like the cochlea.

The MEMS die is fabricated using the microfabrication processes detailed in the [supplementary material](#) and Zhao.¹⁸ Seven radii were fabricated ranging from 100 to $275 \mu\text{m}$ increasing in $25 \mu\text{m}$ increments. Theory indicates that the larger diaphragms will perform better, but due to manufacturing difficulties, $175 \mu\text{m}$ is the largest radius we were able to reliably produce. The average measured capacitance for this diaphragm radius is $3.78 \pm 0.17 \text{ pF}$ measured for 11 devices at 1 kHz with a Keysight E4980A Precision LCR Meter. This differs from the expected capacitance (2.7 pF), indicating approximately 1.1 pF of stray capacitance on the die.

A sample $175 \mu\text{m}$ radius one-electrode device is shown in Fig. 1(c). For characterization and testing, the transducer was mounted on a printed circuit board (PCB) with 3M scotch double-sided tape followed by cyanoacrylate (Krazy Glue) to seal the back cavity. The transducer was wire-bonded from the electrode pads to the PCB (Protoconnect LLC, Ann Arbor, MI). Silicone encapsulates the bond pads, wires, and bonding joints to ensure electrical integrity, but the diaphragm remains exposed [see Fig. 1(d)].

Experimentation begins with voltage actuation of the sensor before connection to the buffer circuit, which allows us to verify transducer functionality, estimate residual stress in the AlN film, and determine d_{31} of the AlN.¹¹ Voltage actuation experiments for piezoelectric cantilever beams have been described previously.^{19–22} For the experiments discussed here, 1 V is applied to the electrode pads and the

deflection is measured at the center of the circular diaphragm with a laser Doppler vibrometer (LDV). A droplet of water was placed on the PCB encapsulating the sensor to complete the in-water actuation testing. Figure 2(a) includes the model and measurement comparison of a pure tone sinusoidal frequency sweep from 300 to 600 kHz. Using an AlN compliance, s_{11}^E , equal to $2.9 \times 10^{-12} \text{ m}^2/\text{N}$ in the FEA simulation,²³ the diaphragm stress is adjusted to align the modeled resonant frequency with the frequency indicated in the experimental data. In order to achieve an in-air structural resonance of 552 kHz, we estimate that an average value of 250 MPa tensile stress is present in the AlN layer of the sensor diaphragm. This value was similar to the average measured wafer-level AlN residual stress (approximately 350 MPa, measured by the manufacturer), a difference that is likely due to the non-uniform distribution of residual stress across the area of the wafer. We confirmed that the small displacement fluctuation around 343 kHz in both the measured and modeled results is caused by the acoustic resonance of the back cavity, indicating a good acoustic seal (see also Zhao¹⁸).

Additional actuation tests from 1 to 200 kHz were performed in air and water. The in-air low-frequency displacement sensitivity of 0.39 nm/V matches predictions when $d_{31} = -0.67 \text{ pm/V}$. This value is low compared to the reported values^{11,12,17,24,25} of -0.9 to -2.8 pm/V , suggesting poor AlN film quality. With the adjusted d_{31} and residual stress included, the properties of the fluid environment are changed to model in-water actuation. The low-frequency displacement response is again 0.39 nm/V, and the resonant frequency is 133 kHz. Figure 2(b) shows the matched model and data in air and water, indicating accurate modeling of the fluid-loaded structure.

The method and equipment details for the air and water acoustic sensing configurations are provided in the [supplementary material](#). Briefly, in-air sensing tests are conducted by sending a chirp signal to a speaker connected to a PVC pipe where the transducer and a reference microphone (2520, Larson Davis, USA) are placed. In-water testing is conducted using the vibrating column of the water method described by Schloss and Strasberg²⁶ with a 3 ml syringe (3 ml Global Easy Glide Disposable Syringe, Dispense All, USA) glued to the PCB. Electrical components that buffer and amplify the voltage signal from the transducer are soldered to the PCB. Additional description of this circuit is also given in the [supplementary material](#).

FIG. 2. Actuation results. (a) Modeled (dashed line) and measured (solid line) frequency sweep of the transducer in air from 300 to 600 kHz. The in-air structural resonance is 552 kHz. (b) Amplitude and phase of the modeled (dashed) and measured (solid) frequency response of the transducer both in air and in water from 1 to 200 kHz. The in-water structural resonance is 133 kHz, and the low-frequency displacement in both fluids is 0.39 nm/V. The model and measurements match in both air and water.

The frequency sensitivity, measured in air and water, is compared to the modeled result from 10 Hz to 1 kHz in Fig. 3(a). A single dashed line for the model describes the expected constant low frequency sensitivity for air and water loading. In-air tests (blue line) are restricted to below 1 kHz due to resonant modes in the PVC pipe above that frequency. The water tests (red line) are valid below approximately 100 Hz since a resonance corresponding to the cantilevered syringe is observed around 200 Hz. In-water data are plotted to 1 kHz for completeness. The low frequency sensitivity referred to the input of the amplifier (i.e., not including 33 dB amplifier gain) is $-114.5 \text{ dB re } 1 \text{ V/Pa}$ ($1.87 \mu\text{V/Pa}$). The FEA model, including residual stress, predicts a sensitivity of $7.5 \mu\text{V/Pa}$. Stray capacitance in the system creates a voltage divider that reduces the sensitivity to $1.87 \mu\text{V/Pa}$. The calculated total stray capacitance in the system is approximately 8 pF.

The in-air linearity of the output voltage to input pressure is tested from 15 to 630 Pa at 1 kHz. The data points and linear curve fit are plotted in Fig. 3(b), also without amplifier gain included. A slope of $1.85 \mu\text{V/Pa}$ was obtained (with $0.017 \mu\text{V/Pa}$ standard error), which agrees with the sensitivity as measured from the frequency response within 2% error. Underwater testing indicated survivability up to 9.4 kPa, an upper pressure limited by the shaker amplitude. Simulation results predict that the diaphragm transducer will not break within 200 kPa excitation even with a 250 MPa residual stress; however, this is untested.

The electrical noise due to the amplifier circuit is also needed to calculate the MDP of the sensor. The measured input referred noise spectral density is plotted in gray in Fig. 4. During data collection, the transducer and amplifier are shielded in a grounded metal box and placed in an acoustic chamber; data were acquired at a 500 kHz sampling rate with no anti-aliasing filter. The black dash-dotted line is the total calculated input referred noise (IRN). It is made up of contributions from the JFET common source amplifier (red, green, and magenta lines) and the dielectric losses of the piezoelectric material in the sensor (blue line). As seen in the figure, between approximately 200 Hz and 1 kHz, the primary noise source is the sensor noise. To achieve a modeled result that matches the collected data, a value of 0.7% is found for $\tan\delta$ of the AlN. Above 1 kHz, the presence of aliased high-frequency white noise increases the measured noise as compared to the model-predicted level. Using the noise data collected here in $\mu\text{V}/\sqrt{\text{Hz}}$ divided by the sensitivity data from Fig. 3(a) in $\mu\text{V/Pa}$, the

FIG. 3. Sensing results. (a) Amplitude and phase of the modeled (dashed) and measured (solid, without amplifier gain) in-air and in-water frequency sensitivity of the transducer from 10 to 1000 Hz. The low-frequency sensitivity is $-114.5 \text{ dB re } 1 \text{ V/Pa}$. Water data are valid below 100 Hz. (b) In-air linearity data (open circles) and the linear regression fit (dashed). The slope of the regression line ($1.85 \mu\text{V/P}$) confirms the measured frequency sensitivity, again without amplification.

FIG. 4. Measured input referred noise spectrum (gray line) and total calculated input referred noise (dash-dot black line) from 0.01 to 100 kHz. Contributions of circuit noise sources such as JFET drain current IRN (red), JFET gate current IRN (green), and load resistor IRN (magenta) are included in the total modeled noise. Deviation of data above 1 kHz is due to aliased high-frequency noise. Based on the fit to data from 200 Hz to 1 kHz, $\tan \delta$ is 0.7%.

calculated MDP in a frequency-centered 1 Hz bandwidth is 43.7 mPa (67 dB SPL) at 100 Hz and 10.9 mPa (55 dB SPL) at 1 kHz.

To compare the performance of our sensor with that of other piezoelectric diaphragms in the literature, we consider the MDP with respect to the diaphragm area (A_d) at 100 Hz in Fig. 5. First, the MDP of the experimentally tested device ($A_d = 9.6 \times 10^{-8} \text{ m}^2$) is indicated by a red “+.” We have drawn a red dashed line through this point along a slope of $A_d^{-1.5}$. As shown in Eq. (1) and validated by our FEA model, MDP follows $A_d^{-1.5}$ for a fixed r_2/r_1 ratio. Hence, this dashed line represents the MDP of our fabrication if we increase the area. The dotted line intersecting the red “+” is the result of connecting multiple devices in parallel, shown to decrease MDP proportional to $A_d^{-0.5}$ since doubling the area doubles the sensitivity while the noise rises by $\sqrt{2}$. The end point of the dotted line labeled “4” indicates the MDP for 4 identical sensors connected in parallel, provided as a reference point.

Next, three other piezoelectric diaphragm transducers from the literature^{1,8,27} are each marked with a black “+.” The diaphragm area and MDP were determined from sensor dimensions and measured data provided in these articles. If we solely compare the MDP values, it

FIG. 5. MDP results at 100 Hz. The tested sensor is marked with a red “+” and a theoretical sensor with decreased residual stress and improved AlN properties is marked with a blue “*.” Comparable piezoelectric diaphragm sensors^{1,8,27} are each marked with a black “+.” Dashed and dotted lines represent the $A_d^{-1.5}$ and $A_d^{-0.5}$ area trend lines, respectively. Points marked with “4,” “9,” “16,” and “25” along the dotted lines indicate how many devices are connected in parallel at each point.

is clear that the performance of sensors described in these references exceeds what we achieve. However, since the diaphragm area plays a significant role in the final MDP, we consider equivalent area comparisons. The red dashed line indicates that increasing the diaphragm area of our single sensor would result in an MDP that is lower than that of other sensors in the literature with the same area. This can, in part, be attributed to the through-thickness composition of the diaphragm. The referenced sensors have thicker diaphragms than our design ($2.1 \mu\text{m}$ vs $2.6 \mu\text{m}$,¹ $3.7 \mu\text{m}$,⁸ and $18 \mu\text{m}$,²⁷) which increases the diaphragm stiffness and decreases sensitivity. Different material choices also play a role in determining diaphragm stiffness for these sensors. Additionally, as we have seen in our fabrication process, the residual stress in the diaphragms will affect the sensitivity. Finally, as indicated by the trend of the red dotted line, we see that connecting several as-fabricated sensors in parallel does not sufficiently decrease the MDP to compete with the other sensors.

The performance of our sensor suffers due to the residual stress in the diaphragm and the poor AlN film quality ($d_{31} = -0.67 \text{ pm/V}$). Although it is impossible to completely eliminate residual stress in the AlN, lower residual stress is possible.¹⁵ The blue “*” represents the predicted MDP for a $9.6 \times 10^{-8} \text{ m}^2$ area sensor with low stress (50 MPa) and high-quality piezoelectric film ($d_{31} = -2.51 \text{ pm/V}$ and $\tan \delta = 0.002$).¹⁷ Consistent with the red lines above, the blue dashed and dotted lines that intersect this point in Fig. 5 represent $A_d^{-1.5}$ and $A_d^{-0.5}$ trends, respectively. Sensors of equivalent diaphragm area, decreased residual stress, and improved AlN properties achieve MDP values more than 11 dB lower than the other piezoelectric diaphragm sensors in the literature.¹

To conclude, this paper demonstrates the design, fabrication, and testing of a MEMS piezoelectric diaphragm transducer. Of the fabricated devices, a functional $175 \mu\text{m}$ radius sensor was tested and is shown to work in air and water with the same sensitivity. A key contribution of this work is the set of design rules summarized in Fig. 5. We confirm that the predicted trend for increasing the diaphragm area [Eq. (1)] also holds for diaphragms with residual stress. Additionally, we see that a single larger diaphragm more effectively decreases the MDP than multiple parallel-connected sensors. However, if the maximum radius is dictated by maintaining a high resonant frequency and large bandwidth for an application, then parallel-connected devices are still beneficial. Finally, the experimental MDP was higher than predicted due to high residual stress in the diaphragm and the poor piezoelectric coefficient of the AlN. However, with improved processing to control these variables, this work demonstrates that small sensors with improved resolution (lower MDP) and large dynamic range can be fabricated to satisfy the requirements of various applications.

See the [supplementary material](#) for additional modeled device parameters as well as fabrication process details, the amplifier circuit, in-air sensing experimental setup, and in-water experimental setup.

AUTHORS' CONTRIBUTIONS

A.E.H. and C.Z. contributed equally to this work.

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Nanofabrication Facility at the University of Michigan who helped us while we fabricated the devices in the lab.

The authors declare the following conflicts of interest: KG as the co-founder of Vesper Technologies, a piezoelectric sensor company; LP as the co-founder of Singular Medical, a company whose medical devices utilize sensors.

DATA AVAILABILITY

The data that support the findings of this study are available from the corresponding author upon reasonable request.

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