Modeling, Fabrication, and Characterization of Piezoelectric Micromachined Ultrasonic Transducer Arrays Based on Cavity SOI Wafers

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Abstract—This paper presents high fill-factor piezoelectric micromachined ultrasonic transducer (PMUT) arrays fabricated via a novel process using cavity SOI wafers. The simple three-mask fabrication process enables smaller diameter PMUTs (25 μm) and finer pitch than previous processes requiring through-wafer etching. PMUTs were fabricated with diameters from 25 to 50 μm, resulting in center frequencies from 13 to 55 MHz in air. Two types of devices, having different piezoelectric layers, lead zirconium titanate (PZT), and aluminum nitride (AlN), were fabricated and characterized. Comparing 50-μm diameter devices, the PZT PMUTs show large dynamic displacement sensitivity of 316 nm/V at 11 MHz in air, which is ∼20% higher than that of the AlN PMUTs. Electrical impedance measurements of the PZT PMUTs show high electromechanical coupling $k_p^2 = 12.5\%$ and 50-Ω electrical impedance that is well-matched to typical interface circuits. Immersion tests were conducted on PZT PMUT arrays. The fluid-immersed acoustic pressure generated by an unfocused 9 × 9 array of 40-μm diameter, 10-MHz PZT PMUTs, measured with a needle hydrophone 1.2 mm away from the array, was 58 kPa with a 25 Vpp input. Beam forming based on electronic phase control produced a narrow, 150-μm diameter, focused beam over a depth of focus >0.2 mm and increased the pressure to 450 kPa with 18 Vpp input. [2014-0324]

Index Terms—PMUT, piezoelectric, ultrasound transducers, micromachined transducers, PZT, AlN, cavity SOI.

I. INTRODUCTION

ULTRASONIC transducers have been used in many applications, such as nondestructive testing (NDT), ranging and velocity sensing, industrial automation, object recognition, collision avoidance, and medical imaging [1]–[5]. Conventional ultrasonic transducers are largely based on bulk piezoelectric ceramic with poor acoustic coupling to air or water, and are also expensive to machine into two-dimensional (2D) transducer arrays needed for 3D imaging [6], [7]. In contrast, micromachined ultrasonic transducers (MUTs) have a compliant membrane structure with low acoustic impedance for good coupling to air and liquids [8]–[10]. Furthermore MUTs have several other advantages over conventional ultrasonic transducers, including small element size, low power consumption, improved bandwidth, low cost, easy fabrication of large arrays with compact designs, and easy integration with supporting electronics [11]–[14].

MUTs can be divided into two types based on the actuation mechanism: capacitive MUT (CMUT) [15] and piezoelectric MUT (PMUT) [2], [16]. Compared with well-developed CMUTs, PMUTs do not require a high polarization voltage (which can exceed 190 V for CMUTs [17]), to achieve the required transducer sensitivity. While the required polarization voltage of a CMUT diminishes as the capacitive gap is decreased, small gaps require tight fabrication tolerances and can result in reduced manufacturing yield. PMUTs also have the advantage of higher capacitance, which results in lower electrical impedance, allowing better matching to supporting electronic circuits and less sensitivity to parasitic capacitance.

When designing MUT arrays, it is desirable to minimize the spacing between MUTs because this results in high fill-factor, and therefore greater acoustic efficiency per unit area, and because it enables the MUTs to be spaced with a pitch that is equal to or less than half the acoustic wavelength (λ/2), thereby minimizing grating lobes [18], [19]. Front-side etching using a sacrificial layer and etch holes has been used to make a high fill-factor PMUT array, but required a complicated multi-layer fabrication process and an additional layer to seal the etch holes after the release etch [2]. Through-wafer etching is a simpler process, but it results in a large PMUT dimension and pitch. Compared with wet isotropic etching [6], through-wafer DRIE (deep reactive ion etching) improves fill-factor but the PMUT dimension and pitch are still limited by the wafer thickness and DRIE aspect ratio. For example, PMUTs with 65 μm dimension and 100 μm pitch were fabricated via through-wafer DRIE [10]. A second challenge for small-dimension PMUTs fabricated via through-wafer etching is the PMUT’s center frequency is increasingly sensitive to diameter variations resulting from DRIE tolerances. To solve this issue, a two-step DRIE process was proposed, allowing 50 μm dimension and 80 μm pitch PMUTs to be fabricated [20]. Here, we avoid the need for through-wafer etching, relying on a more manufacturable process based on cavity SOI wafers.
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Fig. 1. (a) Cross section of a single PMUT, and out of plane deflection of the PMUT with (b) positive and (c) negative voltage inputs when functioning as a transmitter.

with precisely defined cavity diameter to produce PMUTs with a minimum diameter of 25 μm and minimum pitch of 45 μm. The process is both simple, requiring only 3 photomask layers, and high performance, enabling roughly 4× more PMUTs per unit area than previous through-wafer DRIE processes.

II. PMUT DESIGN AND MODELING

A. PMUT Working Principle

Each PMUT functions as both a transmitter and receiver. As a transmitter, the electric field between the top electrode (TE) and the bottom electrode (BE) creates a transverse stress in the piezoelectric layer due to the inverse piezoelectric effect. The generated stress causes a bending moment which forces the membrane to deflect out of plane, launching an acoustic pressure wave into the surrounding medium, as shown in Figure 1. As a receiver, an incident pressure wave deflecting the plate creates transverse stress which results in charge on the electrodes due to the direct piezoelectric effect.

A 3D schematic diagram of a PMUT array based on cavity SOI wafers and the cross section of a single PMUT are shown in Figure 2. In the 72 × 9 array, the top electrodes of the 9 PMUTs in each column are connected together to minimize the number of electrical connections to the die. The PMUTs demonstrated here are fabricated using cavity SOI wafers with a 2.5 μm device Si layer over 2 μm deep vacuum-sealed cavities of various diameters. Devices were fabricated using both lead zirconium titanate (PZT) and aluminum nitride (AlN) piezoelectric layers to allow the performance of these two types of PMUTs to be compared.

Fig. 2. (a) 3D schematic diagram of a PMUT array based on cavity SOI wafers; (b) optical images of the fabricated 72 × 9 PMUT array.

Fig. 3. FEM simulated electrical potential in the middle of the piezoelectric layer at the center of the membrane with various thickness of the piezoelectric layer on a fixed 2.5 μm Si layer under a uniform 100 kPa pressure.

B. Film Thickness Optimization

Finite element method (FEM) simulation was used to determine the best piezo/silicon layer stack to optimize receiving performance. Figure 3 shows electrical potential in the middle of the piezoelectric layer at the center of the membrane with various thicknesses of the piezoelectric layer on a fixed 2.5 μm Si layer under a uniform 100 kPa pressure. The optimal designs for PZT and AlN PMUTs are 1.1 μm PZT/2.5 μm Si and 0.8 μm AlN/2.5 μm Si, respectively. AlN PMUTs show ∼3× higher optimal receiving sensitivity due to its small dielectric constant, while PZT has a larger piezoelectric coefficient $e_{31}$ than that of AlN, and therefore higher transmitting sensitivity.
Unlike conventional bulk piezoelectric ultrasonic transducers, whose resonant frequency is only defined by the transducer thickness, a PMUT’s resonant frequency is defined by both the membrane thickness and diameter. When the PMUT diameter is small relative to the acoustic wavelength, the beam pattern is more omnidirectional, which enables a wide field of view. Mathematically, the beam pattern is modeled using the far-field directivity, which is given by [18]:

$$D_{dir}(\theta) = \frac{48 J_2(ka \sin \theta)}{(ka \sin \theta)^3}$$

where \(\theta\) is the angle of incidence, \(a\) is the PMUT radius, \(k = 2\pi / \lambda\) is the wavenumber, and \(J_2\) is the Bessel function of the third kind. The theoretical directivity for PMUTs operating at 10 MHz (\(\lambda = 150 \mu\text{m} \text{ in tissue}\)) with 30 \(\mu\text{m}\), 100 \(\mu\text{m}\), and 200 \(\mu\text{m}\) diameter is shown in Figure 4, demonstrating that the 30 \(\mu\text{m}\) PMUT produces the most omnidirectional pattern.

C. Electrode Area Optimization

The out of plane deflection of an edge-clamped circular plate with uniform pressure load, \(p\), is [21]:

$$w(r) = \frac{p(a^2 - r^2)}{64D}$$

where \(r\) is the radial coordinate, \(a\) is the plate radius, and \(D\) is the flexural rigidity. The mean stress in the radial direction, \(\sigma_r(r)\), and tangential direction, \(\sigma_\theta(r)\), can be obtained from:

$$\sigma_r(r) = -\frac{E z_p}{1 - \nu^2} \left( \frac{\partial w(r)}{r \partial r} + \nu \frac{\partial w(r)}{\partial r} \right)$$

$$\sigma_\theta(r) = -\frac{E z_p}{1 - \nu^2} \left( \frac{1}{r} \frac{\partial w(r)}{\partial r} + \nu \frac{\partial^2 w(r)}{\partial r^2} \right)$$

where \(z_p\) is the distance from the middle of the piezoelectric layer to the neutral axis of deflection. Substituting (1) into (3) and (4), we obtain:

$$\sigma_r(r) = -\frac{Ep z_p}{16D(1 - \nu^2)} \left( (1 + \nu) a^2 - (3 + \nu) r^2 \right)$$

$$\sigma_\theta(r) = -\frac{Ep z_p}{16D(1 - \nu^2)} \left( (1 + \nu) a^2 - (1 + 3\nu) r^2 \right)$$

To maximize the electromechanical coupling, the top electrode should cover an area where there is no sign change of the stress. Because stresses in both the radial and tangential directions, \(\sigma_r(r)\) and \(\sigma_\theta(r)\), contribute piezoelectric charge, the location where \(\sigma_r(r) + \sigma_\theta(r) = 0\) is used to determine the edge of the top electrode. Figure 5 shows stresses in the radial and tangential directions for a 50 \(\mu\text{m}\) diameter AlN PMUT with 1 Pa uniform pressure load on its surface. The radius where the sum of stresses is zero, \(r = \sqrt{2a}/2\), is the optimal electrode radius in receiving mode. An analytical study on electrode optimization of PMUTs with different boundary conditions was reported in [22], where clamped and simply-supported plates show maximum transmitting efficiency when the electrode radius is 60% and 100% of the plate radius, respectively. An analytical study on PMUTs with multiple electrodes was reported in [23].

III. FABRICATION

PMUT arrays were fabricated via a simple 3-mask process. A cross-section of the fabrication process flow is shown in Figure 6. The process is based on custom-fabricated cavity SOI wafers (IceMos Technology, Belfast, Ireland), as shown in step (a). The first mask is used to pattern 2 \(\mu\text{m}\) deep cavities in the handle wafer, after which the handle and device wafers are bonded in vacuum, followed by grinding and polishing to produce the desired 2.5 \(\mu\text{m}\) thickness of the device layer Si. The cavities define the location of each PMUT and are vacuum-sealed to eliminate the possibility of squeeze-film damping beneath the PMUT membrane. Alignment marks etched into the handle wafer at the same time as the cavities are exposed by selectively etching openings in the Si device layer. The bottom electrode metal and piezoelectric layer are then deposited via sputtering. In AlN PMUTs, sputtering is conducted at <400 °C and begins with a 25 nm AlN seed layer, followed by a 100 nm Mo layer and the 800 nm AlN layer. The PZT sputtering process is conducted at 600 °C and a 20 nm conductive oxide layer SrRuO\(_3\) (SRO) is sputtered between the PZT layer and the 100 nm Pt bottom electrode to improve lifetime by reducing oxygen defects accumulating at the Pt/PZT interface when the polarization is switched [24].
Fig. 6. Fabrication process flow of PMUT array based on cavity SOI wafers. a) Customized cavity SOI wafer. b) Bottom electrode metal layer, piezoelectric layer deposition, and via etching. c) Top electrode lift-off.

After sputtering, the second mask is used to define vias to the bottom electrode through wet etching, step (b), with HNO₃/BHF/H₂O in a volumetric ratio of 5/5/90 at room temperature used to etch PZT [25], and TMAH-based developer (MF-319, Rohm & Haas Electronic Materials) at 60 °C used to etch AlN. In the final step, step (c), the third mask is used to define the top electrode through lift-off of 150 nm of Al deposited by sputtering. Devices were fabricated with 25 μm, 30 μm, 40 μm and 50 μm diameters, corresponding to theoretical fundamental mode frequency ranging from 13 MHz to 55 MHz in air. A 20 μm minimum space was used between PMUTs, producing a minimum pitch of 45 μm for 25 μm diameter PMUTs. Scanning electron microscope (SEM) images of the cross section of a single PMUT are shown in Figure 7. Figure 8(a) shows X-ray diffraction (XRD) measurement of the PZT film on Pt bottom electrode, where the (100) peak is much larger than the (110) peak. Figure 8(b) shows rocking curve measurement with 1.3° full width at half magnitude (FWHM). A relatively large (100) peak and a small 1.3° FWHM, demonstrate the good deposition of the PZT film and predict a high piezoelectric coefficient.

IV. CHARACTERIZATION

The frequency response of PZT and AlN PMUTs were measured in air via a laser dropper vibrometer (LDV, OFV 512 and OFV 2700, Polytec) in conjunction with a network analyzer (E5061B, Agilent Technologies) as shown in Figure 9. The PZT PMUT shows a large dynamic displacement sensitivity of 316 nm/V, ~20× higher than that of AlN PMUTs with the same 50 μm diameter. When tested in air and in vacuum, the two PMUT types have slightly different quality factors (Q = 140 for AlN and Q = 115 for PZT PMUTs with 50 μm diameter) but the difference is not

![Image](image_url)
due to material properties. Instead, the $Q$ in air and in vacuum is determined by anchor loss and the difference between the two device types is caused by the difference in piezoelectric film thickness (0.8 $\mu$m AlN and 1.1 $\mu$m PZT), as expected from theory [26], [27]:

$$Q_{\text{anchor}} \propto (a/t)^3$$  \hspace{1cm} (7)

where $Q_{\text{anchor}}$, $a$, and $t$ are the anchor-loss limited $Q$, radius, and total membrane thickness respectively.

FEM and measured results of the center frequencies in air of both PZT and AlN PMUTs with various diameters are shown in Figure 10(a). The maximum measured frequency was limited by the 28 MHz bandwidth of the LDV. The measurement results agree well with FEM results, indicating the FEM model (using COMSOL MultiPhysics) can be used to efficiently guide PMUT design. The theoretical frequency of the PMUT’s fundamental vibration mode in air can be obtained from:

$$f_n = \sqrt{\frac{(3.2/a)^3 D}{\rho}} \frac{1}{2\pi}$$  \hspace{1cm} (8)

where $a$ is the radius, $\rho$ is area mass density and $D$ is the flexural rigidity of the plate. $D$ can be obtained by integrating from the bottom surface of the Si device layer to the top electrode:

$$D = \int_{\text{Bottom}}^{\text{Top}} \frac{E(z)z^2}{1-\nu(z)^2} dz$$  \hspace{1cm} (9)

where $E(z)$ is the Young’s Modulus and $\nu(z)$ is the Poisson’s ratio of the material at a distance $z$ from the neutral axis. Referring to (8), because PZT has a smaller Young’s modulus ($E_{\text{PZT}} = 76$ GPa, $E_{\text{AlN}} = 330$ GPa) and larger density ($\rho_{\text{PZT}} = 7.7$ g/cm$^3$, $\rho_{\text{AlN}} = 3.2$ g/cm$^3$) than that of AlN, PZT PMUTs show a ~40% lower resonant frequency than that of AlN PMUTs of the same diameter.

Two metrics for PMUT performance are the displacement sensitivity $d_s(f_n)$, and fractional bandwidth, $\Delta f/f_n$. The peak displacement sensitivity occurs at the resonance frequency $f_n$ and is equal to the static (zero-frequency) displacement sensitivity multiplied by the quality factor:

$$d_s(f_n) = Q d_s(0)$$  \hspace{1cm} (10)

Because the fractional bandwidth is inversely proportional to the quality factor, $\Delta f/f_n = Q^{-1}$, the static displacement sensitivity is also equal to peak displacement-bandwidth product:

$$d_s(0) = d_s(f_n) \cdot \Delta f/f_n$$  \hspace{1cm} (11)

Measurements in air, where $Q$ is high, result in large peak displacement and narrow bandwidth, while measurements in fluid, where $Q$ is near unity, result in much smaller peak displacement and wider bandwidth. For this reason, the static displacement sensitivity, which is independent of $Q$, provides a better measure of performance than the peak displacement sensitivity does.

In terms of the design parameters, the static displacement sensitivity can be expressed as [28]:

$$d_s(0) \propto e_{31} Z_n / D$$  \hspace{1cm} (12)

where $e_{31}$ is the piezoelectric coefficient, and $Z_n$ is the distance from the middle of the piezoelectric layer to the neutral axis [28]. Figure 10(b) shows FEM and measured results of static displacement in air for both PZT and AlN PMUTs with various diameters. Overall, the displacement sensitivity scales roughly with the inverse of $f_n$. For the same diameter, 50 $\mu$m, the PZT PMUT shows a ~28x higher static displacement than that of the AlN PMUT. Piezoelectric coefficients were extracted from these measurements, yielding $e_{31} = -8$ C/m$^2$ for PZT and $-0.5$ C/m$^2$ for AlN, in good agreement with published values [29]–[31].


Figure 11 shows the displacement frequency response measured in air of 50 μm PZT PMUTs before and after poling at 20 V DC for 1 min at room temperature. The poling improves the piezoelectric coefficient $e_{31}$ by $\sim 100\%$ (from $-4 \, \text{C/m}^2$ to $-8 \, \text{C/m}^2$) but unlike PZT films made via the Sol-gel method [32], the sputtered PZT film has a large initial piezoelectric coefficient before poling.

Impedance measurements of individual PZT and AlN PMUTs with 50 μm diameter are shown in Figure 12(a) and (b), respectively. The resonant frequencies extracted from the impedance measurement agree well with the values extracted from LDV measurements. The electromechanical coupling coefficient, $k_t^2$, can be extracted from these impedance measurements using [33]:

$$k_t^2 = \frac{f_a^2 - f_r^2}{f_a^2} = \frac{C_{\text{MEMS}}}{C_0 + C_{\text{MEMS}}}$$  \hspace{1cm} (13)

where $f_a$ and $f_r$ are the anti-resonant frequency and resonant frequency, respectively, and $C_{\text{MEMS}}$ and $C_0$ are the motional capacitance and the electrical capacitance, respectively. From impedance measurements, the dielectric constants of the PZT and AlN films are extracted to be $\varepsilon_{33, \text{PZT}} = 528$ and $\varepsilon_{33, \text{AlN}} = 9$, respectively. Parasitic capacitance $C_p$ caused by the bond-pad and interconnect is calculated to be 74 pF based on the measured dielectric constant and metal area. After calibration to remove $C_p$, $C_{\text{MEMS}} = 0.6 \, \text{pF}$ and $C_0 = 4.17 \, \text{pF}$ are extracted from measured electrical impedance of the PZT PMUT, Figure 12(a), resulting in $k_t^2 = 12.5\%$.

In addition, the PZT PMUT has 50 Ω impedance, which is matched to the impedance of conventional remote readout electronics and coaxial cables. However, PZT’s high dielectric constant results in large parasitic $C_p$. $C_p$ can be reduced by removing PZT from beneath the interconnect and bond pads, by adding an additional dielectric layer, or both.

For medical imaging applications, a biocompatible layer is frequently used to encapsulate MUTs. The biocompatible layer provides insulation and increases the lifetime of devices in a fluid environment. Polydimethylsiloxane (PDMS) is used in this work as it is biocompatible and has similar acoustic impedance to that of human tissue, 1.5 Mrayl. To demonstrate that the PDMS layer does not significantly reduce the array’s output pressure, acoustic pressure measurements were conducted on a $9 \times 9$ array of 40 μm PZT PMUTs with and without a 250 μm PDMS sheet on the PMUT surface. Fluorinert FC-70 (3M) was used for fluid immersed measurement because it has acoustic impedance similar to that of human tissue and high electrical resistivity which eliminates the need for epoxy protection of the bond-wires. Four cycles of 25 Vpp short-duration (0.4 μs) pulses were used to drive the 10 MHz PMUT array and a 40 μm diameter needle hydrophone (Precision Acoustics) measured the resulting ultrasound pressure. In this measurement, the PMUTs were driven without individual phase control, resulting in an unfocused acoustic beam with 58 kPa peak-to-peak amplitude at 1.2 mm away from the array, Figure 13. The fast Fourier transform (FFT) of the received pressure signal shows a 3.1 MHz bandwidth at 10.4 MHz center frequency. The measured pressure is similar both with and without the PDMS sheet. However, a small difference in the time-of-arrival arises due to the difference in the speed of sound in PDMS and fluid, and a small intensity reduction is
Fig. 13. The acoustic pressure generated by PMUTs excited with a 4-cycle 10 MHz 25 Vpp input, demonstrating that a 250 μm thick PDMS layer has minimal effect on the pressure amplitude.

Fig. 14. Measured pressure map of a 15 × 9 array driven at 10 MHz with 18 Vpp using phased-array beam forming.

caused by standing-wave interference from the PDMS-fluid interface.

To obtain a higher acoustic pressure over a narrow focused volume, the PMUT array is electronically focused via the beam-forming (phased-array) method wherein the voltage signals applied to the PMUTs in the array are delayed by specific phase shifts. Acoustic lenses, commonly used to focus conventional ultrasonic transducers, cause fabrication complexity and have a fixed focal length. Unlike an acoustic lens, phased array focusing can achieve variable focal length by changing the phase delay between adjacent PMUTs. Furthermore a phased array focusing can achieve variable focal length by changing the phase delay between adjacent PMUTs. The needle hydrophone was mechanically scanned with 50 μm step size to obtain the measured acoustic pressure map shown in Figure 14. The pressure map was generated by a 15 × 9 PZT PMUT array (40 μm diameter and 60 μm pitch) excited with 18 Vpp. The PMUTs are connected in columns (as shown in Figure 2(b)), each driven with a selected phase delay using an ultrasound transmitting evaluation kit (Texas Instruments, Dallas, Texas). The selected phase delay is calculated based on the acoustic path length from each PMUT to the desired focus point, ensuring that pulses launched from each PMUT arrive simultaneously at the focus. The measurement results show 450 kPa peak-to-peak amplitude at the focus point, a beam width of 150 μm FWHM, and a depth of focus >0.2 mm. Pressure measurements with and without beam-forming demonstrate a 3× increase in pressure using beam-forming, as shown in Figure 15.

V. CONCLUSION

This paper presented PMUTs fabricated via a simple three-mask process using cavity SOI wafers. The process enables the fabrication of PMUTs with small diameter (25–50 μm) and fine pitch (45–70 μm), over a range of frequencies from 13 MHz to 55 MHz. Unlike previous processes that use through-wafer DRIE to release the PMUTs, cavity SOI allows smaller spacing between PMUTs, resulting in higher fill factor, and has the practical advantage that the cavity SOI wafer does not have through-wafer holes or require back-side wafer processing, simplifying wafer handling and increasing device yield.

PMUTs were fabricated using both AlN and PZT piezoelectric layers to compare the performance of these two materials. While sputtered AlN has the advantages of lower deposition temperature (400 °C versus 600 °C for PZT) and much lower dielectric constant, sputtered PZT films were shown to have 16 times higher piezoelectric coefficient, ultimately resulting in 28 times higher displacement sensitivity than AlN PMUTs of the same diameter.

Ultrasound experiments conducted using an immersed 15 × 9 PZT PMUT array operating at 10 MHz resulted in 450 kPa peak-to-peak pressure with 18 V peak-to-peak input, a beam width of 150 μm FWHM, and a depth of focus >0.2 mm. These results show that cavity SOI PMUTs are promising devices for ultrasound imaging.

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