Abstract—This paper presents a 1.2-mm diameter high fill-factor array of 1261 piezoelectric micromachined ultrasonic transducers (PMUTs) operating at 18.6 MHz in fluid for intravascular ultrasound imaging. At 1061 transducers/mm², the PMUT array has a 10–20 times higher density than previous PMUT arrays realized to date. Aluminum nitride (AlN)-based PMUTs described in this paper are fabricated using a process compatible with the fabrication of inertial sensors, radio frequency (RF) resonators, and CMOS integrated circuits. The PMUTs are released using a front-side sacrificial etch through etching holes that are subsequently sealed by a thin layer of parylene. Finite element method and analytical results, including resonant frequency, pressure sensitivity, output acoustic pressure, and directivity are given to guide the PMUT design effectively, and are shown to match well with measurement results. Due to the PMUTs thin membrane (750-nm AlN/800-nm SiO₂) and small diameter, a single 25-μm PMUT has approximately omnidirectional directivity and no near-field zone with irregular pressure pattern. PMUTs are characterized in both the frequency and time domains. Measurement results show a large displacement response of 2.5 nm/V at resonance and good frequency matching in air, high center frequency of 18.6 MHz and wide bandwidth of 4.9 MHz, when immersed in fluid. Phased array simulations based on measured PMUT parameters show a tightly focused high-output pressure acoustic beam. [2014-0114]

Index Terms—Aluminum nitride (AIN), piezoelectric, ultrasonic, transducer, micromachined ultrasonic transducer (MUT), piezoelectric MUT (PMUT).

I. INTRODUCTION

NOWADAYS there is a growing interest in ultrasonic transducers, which have been used in many applications, such as nondestructive testing (NDT), ranging and velocity sensing, industrial automation, object recognition, collision avoidance, and particularly medical imaging [1]–[4]. Conventional ultrasonic transducers are largely based on bulk piezoelectric ceramic with poor acoustic coupling to air or liquids, and it is also expensive for them to be machined into two-dimensional (2D) transducer arrays needed for 3D imaging [5], [6]. In contrast, micromachined ultrasonic transducers (MUTs) have a compliant membrane structure with low acoustic impedance for good coupling to air and liquids. 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 [4], [7].

MUTs can be divided into two types based on their actuation mechanism: capacitive MUTs (CMUTs) and piezoelectric MUTs (PMUTs). Compared with well-developed CMUTs, PMUTs do not require a polarization voltage (which can exceed 190 V for CMUTs [8]), to achieve the required transducer sensitivity. This is particularly important for catheter-based ultrasound applications as having a high voltage inside the body requires more complex packaging. In addition, CMUTs require a small gap [9] to achieve a desired sensitivity, which increases complexity and cost of fabrication. Another advantage of PMUTs is that they have higher capacitance, which results in lower electrical impedance, allowing better matching to supporting electronic circuits and less sensitivity to parasitic capacitance. Most of the previous work on PMUTs has focused on lead zirconate titanate (PZT) film, because of its high piezoelectric coefficient. While PZT has higher piezoelectric constants than the aluminum nitride (AIN) film used in this work, the lower dielectric constant (ε₀) of AIN allows for comparable performance to be achieved, especially in terms of sensitivity in the receive mode [10]. In addition, unlike PZT films that require high fabrication temperature (around 800 °C) [11], [12], AIN is deposited at a low temperature (<400 °C) and is a material with full compatibility with CMOS processes [13].

MUTs operating at 5 MHz, where the acoustic wavelength is about 300 μm in tissue, have little advantage over conventional saw-cut piezoelectric ultrasonic transducers. However, saw cutting cannot achieve an element pitch smaller than 100 μm, making micromachining an attractive option for high frequency (>10 MHz) arrays requiring half-wavelength (λ/2) element pitch. Previous PMUTs were fabricated by a through-wafer etching approach [14], [15], resulting in low fill-factor, small element count, and therefore poor acoustic efficiency. Here, we present PMUTs with 10–20× higher density (1061 transducers/mm²) than the highest density PMUT arrays realized to date, 56 transducers/mm² [14] and 123 transducers/mm² [15]. This result was achieved using an AIN fabrication process originally developed for RF MEMS resonators, filters [16] and inertial sensors [17]. The small 5 μm spacing between PMUTs enables a high fill factor, and therefore a large array gain, which is the ratio of the array’s output pressure to that of a single PMUT. In addition, the

Manuscript received April 4, 2014; revised August 31, 2014; accepted September 14, 2014. This work was supported by the Berkeley Sensor and Actuator Center Industrial Members, University of California at Davis, Davis, CA, USA. Subject Editor E.-S. Kim.

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Digital Object Identifier 10.1109/JMEMS.2014.2358991
small array can be used for high resolution medical imaging, especially intravascular ultrasound (IVUS) imaging, which requires small transducer size (<2 mm) and high operating frequency (>10 MHz). In IVUS, close proximity to the target allows the transducer to pick up relatively weak ultrasound signals, which provides diagnostic information inaccessible from a noninvasive transducer.

II. PMUT Array Design

Optical image of the PMUT array demonstrated here and a close-up picture of individual PMUTs are shown in Figure 1. The PMUTs are arranged in annular rings that are electrically connected through the top electrode metal layer. Eight concentric channels are formed by connecting eight groups of adjacent rings. As summarized in Table 1, the number of rings in each channel was selected to yield approximately the same active area (and therefore acoustic pressure output per unit volt) for each channel. As shown in Figure 2, the phase delay of the signal applied to each channel can be controlled in order to control the focus point of the array. This allows a narrow acoustic beam without the need for an acoustic lens and enables electronic focus control, eliminating the need for mechanical scanning along the axis of the acoustic beam.

A. Scaling and Fill-Factor

A PMUT’s resonant frequency is proportional to the membrane thickness, \( h \), and inversely proportional to the radius squared, \( a^2 \) [18]:

\[
f_n \propto \frac{h}{a^2}
\]

(1)

In addition, the pressure sensitivity \( S_p \) is proportional to the radius squared and inversely proportional to the thickness squared,

\[
S_p \propto (a/h)^2
\]

(2)

To compare two PMUTs with the same resonant frequency but different diameters \( a_1 \) and \( a_2 \) using Equation (1) and (2), the ratio of the pressure sensitivities is:

\[
\frac{S_{p1}}{S_{p2}} = \left(\frac{a_1}{a_2}\right)^2 \frac{a_2^2}{a_1^2} = \frac{a_2^2}{a_1^2}
\]

(3)

which means \( S_{p0}^{f0} \), the sensitivity for PMUTs with a fixed frequency \( f_0 \), is inversely proportional to radius squared, \( a^2 \):

\[
S_{p0} \propto 1/ a^2
\]

(4)

For the same area, the PMUT array with a high fill-factor has a large effective area to generate acoustic wave or sense acoustic pressure. The fill-factor of an array with PMUT radius, \( a \), and space between adjacent PMUTs, \( d \), is given by:

\[
FF = \frac{\pi a^2}{(2a + d)^2}.
\]

(5)

Therefore, we can define a figure of merit (FOM), which is the product of the fill-factor, \( FF \), and PMUT sensitivity \( S_{p0}^{f0} \):

\[
FOM = \frac{\pi a^2}{(2a + d)^2} \times \frac{1}{a^2} = \frac{\pi}{(2a + d)^2}
\]

(6)

Eq. (6) shows that for PMUTs with a specific resonant frequency, a smaller PMUT pitch (smaller radius, \( a \), or space between adjacent PMUTs, \( d \)) will give a higher FOM, indicating higher output sound pressure per unit area.

III. SINGLE PMUT DESIGN AND MODELING

A. Vibration Modes

The first and second vibration mode shapes were obtained by FEM, as shown in Figure 3, where the first mode is the best for a high acoustic coupling efficiency. According to
force equilibrium in the z-direction, symmetric vibration of a uniform circular thin plate is described by [18], [19]:

\[ D \nabla^4 (w(r)e^{\text{j} \omega t}) + \rho \frac{\partial^2 (w(r)e^{\text{j} \omega t})}{\partial t^2} = 0 \]  

(7)

where \( w(r) \) is the deflection amplitude of the plate in the z-direction at radial distance \( r \), \( \omega \) is the PMUT working frequency, \( \rho \) is the mass per unit area, and \( D \) is the flexural rigidity of the plate. \( D \) can be obtained from:

\[ D = \int_0^z \frac{E(z)z^2}{1 - v(z)} \, dz \]  

(8)

where \( E(z) \) is Young’s Modulus and \( v(z) \) is Poisson’s ratio of the material at a distance \( z \) from the neutral axis, which is the point where the sum of stresses is equal to zero. The integration is conducted for \( z \) varying from the bottom to the top of the plate. Considering a fully-clamped boundary condition at the circular edge, we can obtain the first mode’s resonant frequency as follow:

\[ f_1 = \sqrt{\frac{3.2/a}{\rho} \frac{D}{2\pi}} \]  

(9)

where \( a \) is the radius of the circular membrane.

Because of increased damping in fluid compared with that in air, the center frequency of the PMUT will be reduced. The center frequency of a clamped circular plate with a fluid load on one side [20]–[22] can be approximately written as:

\[ f_{1, \text{fluid}} = f_1 (1 + 0.67 \rho_{\text{fluid}}/\rho)^{-0.5} \]  

(10)

where \( \rho_{\text{fluid}} \) is the fluid density.

**B. Membrane Deflection**

A cross-section schematic of the PMUT with exaggerated deformation is shown in Figure 4. The PMUT can work as both a transmitter and receiver. As a transmitter, the electric field between the top electrode (TE) and the bottom electrode (BE) creates transverse stress in the AIN piezoelectric layer due to the inverse piezoelectric effect. The generated stress causes a bending moment which forces the membrane to deflect out of plane and launch an acoustic pressure wave into the environment. In the receive mode, an incident pressure wave deflecting the plate creates transverse stress which results in charge between the electrodes due to direct piezoelectric effect.

For analyzing the stress and strain of the bending membrane, the neutral axis of deflection must be determined. Since the membrane vibration amplitude is much smaller than its dimensions, we can assume it to be plane strain, i.e. no strain in the direction vertical to the membrane. As a result, the radial and tangential stresses inside the thin membrane can be obtained using polar coordinate:

\[ \sigma_r = -\frac{E(z)z^2}{1 - v(z)} \left( \frac{\partial^2 w(r)}{\partial r^2} + \frac{1}{r} \frac{\partial w(r)}{\partial r} \right) \]  

(11)

\[ \sigma_\theta = -\frac{E(z)}{1 - v(z)} \left( \frac{1}{r} \frac{\partial w(r)}{\partial r} + \frac{\partial^2 w(r)}{\partial r^2} \right) \]

The \( z \)-coordinate is measured relative to the neutral axis, which is computed from the following equation:

\[ \sum F = \int_{\text{bottom}}^{\text{top}} \sigma_r \, dz = \int_{\text{bottom}}^{\text{top}} \frac{E(z)z}{1 - v(z)} \, dz = 0 \]  

(12)

Given the vibration mode-shape of the membrane, Eq. (11) and (12) allow the stress to be calculated. Using these equations, the layer stack (0.75 \( \mu \)m AIN/ 0.8 \( \mu \)m SiO\(_2\)) was chosen to maximize the piezoelectric stress at a given deflection, thereby maximizing pressure sensitivity. To verify the analytical model, finite element method (FEM) simulation results for the sum of stresses in the radial and tangential directions at the membrane edge \( r = a \) are compared in Fig. 5. Close agreement is observed both for computed stress and for the location of the neutral axis.
C. Equivalent Circuit Model

The equivalent circuit model for a PMUT is shown in Figure 6, where the parameters are as follows: $C_o$ electrical capacitance, $k$ stiffness of the membrane, $b$ mechanical damping, $m$ effective mass, $Z_{ac}$ acoustic impedance, $\eta$ electrical-mechanical coupling coefficient, and $A_e$ effective area, which is equal to $1/3$ the surface area.

Using an energy method [10], [23]–[25], $\eta$ can be obtained:

$$\eta = 18\pi I_m e_{31} Z_p$$  \hspace{1cm} (13)$$

where $e_{31}$ is the piezoelectric coefficient, $Z_p$ is the distance of the middle of the piezoelectric layer from neutral axis, and $I_m$ is an integral representing the piezoelectric coupling to mode shape $w(x)$:

$$I_m = \frac{1}{1-v_{av}} \int_{ra}^{rb} \frac{d^2 w(x)}{dx^2} + \frac{1}{x} \frac{dw(x)}{dx} \frac{dx}{dx}$$  \hspace{1cm} (14)$$

where $v_{av}$ is average Poisson ratio of the PMUT membrane, and $x = r/a$ is the normalized radial coordinate, and $ra$ and $rb$ are the normalized internal and external radius of the top electrode for the PMUT, respectively, as shown in Figure 4. In reality, the PMUT is an edge-clamped membrane, and the membrane deflection [26] can be written using the following expression:

$$w(x) = (1-x^2)^2$$  \hspace{1cm} (15)$$

The stiffness of the PMUT membrane, $k$ can be obtained as:

$$k = 18\pi D I_e / a^2$$  \hspace{1cm} (16)$$

where $D$ is the flexural rigidity of the plate obtained from Eq. (8), and $I_e$ is an integral related to the elastic strain for mode shape $w(x)$:

$$I_e = \frac{1}{1-v} \int_{0}^{1} \left( \frac{d^2 w(x)}{dx^2} - 2 \frac{d w(x)}{dx} \frac{d^2 w(x)}{dx^2} \right) + \left( \frac{d w(x)}{dx} \right)^2 \frac{dx}{dx}$$  \hspace{1cm} (17)$$

D. Comparing the PMUT to a CMUT

Based on the PMUT equivalent circuit model, the equivalent piezoelectric force, $F_p$, can be written as:

$$F_p = V_{ac}\eta = kw_p$$  \hspace{1cm} (18)$$

where $V_{ac}$ is the applied ac voltage, and $w_p$ is PMUT deflection amplitude. Substituting Eq. (13) into Eq. (18), we obtain:

$$\frac{dw_p}{dV_{ac}} = \frac{\eta}{k} = \frac{2\pi I_m e_{31} Z_p}{k}$$  \hspace{1cm} (19)$$

For a first-order comparison, we assume both the PMUT and CMUT work as pistons. The electrostatic force of a CMUT [27], [28] can be obtained as:

$$F_{CMUT} = \frac{C_o (V_{dc} + V_{ac})^2}{2 g_o}$$  \hspace{1cm} (20)$$

where $C_o$ is the capacitance of the CMUT, $g_o$ is the gap distance, and $V_{dc}$ is the polarization voltage. Assuming a lumped-mass system, the electrostatic force can be written:

$$F_{CMUT} = kw_c$$  \hspace{1cm} (21)$$

Replacing Eq. (21) in Eq. (20) and taking the first derivative relative to $V_{ac}$, the transmitting sensitivity of a CMUT can be obtained as:

$$\frac{d w_c}{dV_{ac}} = \frac{C_o (V_{dc} + V_{ac})}{g_o k} \approx \frac{C_o V_{dc}}{g_o k}$$  \hspace{1cm} (22)$$

To compare a PMUT with a CMUT, we assume they have the same stiffness $k$ and membrane radius $a$. To achieve the same sensitivity we equate (19) and (22):

$$\frac{d w_p}{dV_{ac}} = \frac{d w_c}{dV_{ac}}$$  \hspace{1cm} (23)$$

Then, the required DC bias $V_{dc}$ for a CMUT with a given gap $g_o$ can be obtained:

$$V_{dc} = \frac{\eta g_o^2}{\epsilon \pi a^2}$$  \hspace{1cm} (24)$$

The pull-in voltage for the CMUT can be written as:

$$V_{PL} = \sqrt{\frac{8 k g_o}{27 \epsilon_o A}}$$  \hspace{1cm} (25)$$

As shown in Figure 7, either a very small gap (e.g. 200 nm for 50 V DC bias) or a high DC bias (e.g. 5 kV for a 2 µm gap) is required for a CMUT to achieve the same transmitting sensitivity as that of the PMUT proposed here. Additionally, the fact that the required bias voltage is roughly 25% to 50% of the pull-in voltage may lead to reliability issues.
obtained from [18]:

E. Acoustic Pressure Output

The on-axis pressure generated by a moving piston can be obtained from [18]:

\[ P(r, 0) = 2\rho_0 c_0 u_0 \sin \left( \frac{1}{2} k_w r \sqrt{1 + (a/r)^2 - 1} \right) \]  

(26)

where \( c_0 \) is the speed of sound, \( u_0 \) is the vibration velocity amplitude, \( \rho_0 \) is the density, and \( k_w \) is the wave number. When the piston radius is greater than half the wavelength \( (k_w a > \pi) \), the axial pressure displays strong interference effects, and the acoustic field close to the piston is complicated. This irregular near-field pattern is undesired for imaging. When the radius is equal-to or smaller-than half the wavelength \( (k_w a \leq \pi) \), the axial pressure decreases monotonically, approaching an asymptotic \( 1/r \) dependence. The thin-membrane, 25 \( \mu \)m diameter PMUT demonstrated here has a center frequency of 25 MHz in air and 19 MHz in fluid, therefore the PMUT radius is smaller than half the wavelength \( (\lambda = 75 \mu \text{m} \text{at} 20 \text{MHz}) \) in water. As a result, the FEM simulation of the near-field pressure pattern shows that it is extremely uniform.

In reality, the PMUT is an edge-clamped membrane instead of a uniform piston. According to Equation 15, the average velocity of PMUT membrane, \( u_{0av} \), can be obtained [26]:

\[ u_{0av} = u_0/3 \]  

(27)

where \( u_0 \) is the deflection magnitude at the center of the PMUT membrane. This relationship is the reason that the effective area used in the equivalent circuit model (Section III-C) is one third the surface area. The pressure created by the motion of a single PMUT can be obtained [26]:

\[ p = \frac{\rho_0 c_0 u_{0av}}{r} e^{j(wt-kr)} D_{dir}(\theta) \]  

(28)

where \( \theta, D_{dir}(\theta), u_{0av} \) and \( R_0 \) are the angle of incidence, directivity, theoretical surface pressure, and Rayleigh distance, respectively and are defined as follow:

\[ D_{dir}(\theta) = \frac{4\sqrt{3}J_3(ka \sin \theta)}{(ka \sin \theta)^3} \]  

(31)

In the far-field, the PMUT is equivalent to a spherical acoustic source with radius equal to the Rayleigh distance \( R_0 \). Analytical results based on Eq. (31) and FEM simulations of a PMUT with 12.5 \( \mu \)m radius at 20 MHz are shown in Figure 9. These models predict that the PMUT has approximately omnidirectional directivity, which is essential for obtaining a narrow acoustic beam in phased-array beam-forming, as described in the following section.

IV. FABRICATION

The process flow used for device fabrication is shown in Figure 8, where steps (a–d) were performed in the Sandia National Labs AIN MEMS fabrication process [16], [17] and steps (e–f) were performed in the UC Berkeley Marvell NanoLab. This process incorporates a sacrificial polysilicon release pit that precisely defines the PMUT diameter, thereby enabling a small device size (25 \( \mu \)m and even smaller) with close spacing (5 \( \mu \)m) and eliminating the need for through-wafer etching. The sacrificial polysilicon is etched by vapor phase XeF2, releasing the PMUT membranes as shown in step (e). Etching holes are sealed and the device is insulated via vapor-phase deposition of Parylene-C to enable fluid immersion, as shown in step (f). Two kinds of etching holes are designed and fabricated, as shown in Figure 10, (a) 2 \( \mu \text{m} \times 4 \mu \text{m} \) etching holes connected to the PMUTs through a buried polysilicon-filled channel and (b) 1 or 2 \( \mu \text{m} \) diameter etching holes located in the center of each PMUT. Etching holes are required to be small enough so that they can be sealed without filling the cavity beneath the PMUT membrane. Optical images of fabricated PMUTs with the two different etching holes are shown in Figure 10 (c) and (d).

A high fill-factor array (1061 transducers/mm\(^2\)) with etch holes as small as 2 \( \mu \text{m} \times 4 \mu \text{m} \) was successfully released, as shown in Figure 1. A PMUT array with reduced fill factor (155 transducers/mm\(^2\)) allowed successful release using 1 \( \mu \text{m} \) diameter center etch holes, suggesting that XeF2 depletion occurs during the release of dense arrays.
Fig. 10. Cross-section diagram and optical images of two etching hole designs: 2 $\mu$m $\times$ 4 $\mu$m etching holes connected to the PMUT through a buried polysilicon channel (a, c) and 2 $\mu$m diameter etching holes located in the center of the PMUT (b, d).

Fig. 11. Confocal laser microscope measurements showing a 2 $\mu$m by 4 $\mu$m etch hole (a) before and (b) after Parylene sealing and (c) height profile measurements.

3D confocal laser microscope images of an etch hole before and after Parylene sealing are shown in Figure 11, demonstrating that the 3 $\mu$m Parylene layer successfully seals the etch hole. SEM pictures of the cross-section of Parylene sealed (750 nm AlN/40 nm SiO$_2$) PMUTs, and a close-up image of a Parylene-sealed etch hole are shown in Figure 12.

V. CHARACTERIZATION

A Laser Dropper Vibrometer (LDV) (OFV 512 and OFV 2700, Polytec) is used in conjunction with a network analyzer (E5061B, Agilent Technologies) to measure the displacement frequency response in air, as shown in Figure 13.

Fig. 12. SEM images: (a) Cross-section of Parylene sealed PMUT and (b) the close-up picture of buried channel and etching hole.

The average peak displacement sensitivity is 2.5 nm/V at a center frequency of 25 MHz. Measurements of 8 PMUTs selected from each annular ring show a small center frequency mismatch of 0.2 MHz (0.8%) across the array, demonstrating good fabrication uniformity. FEM and measured results of center frequency in air of PMUTs (750 nm AlN/800 nm SiO$_2$) with various diameters are shown in Figure 14. The measured frequencies are $\sim$10% lower than those predicted by FEM simulation. This result could be due to the AlN and SiO$_2$ layers being thinner than...
expected, or may indicate that the FEM boundary conditions (perfectly clamped at the PMUT’s radius) do not capture the true boundary conditions of the PMUT. The frequency response results of a single PMUT in air measured before and after Parylene sealing are shown in Figure 15. The measurements show that the Parylene layer reduces the dynamic displacement sensitivity, equal to the product of static displacement sensitivity and quality factor (Q), from 2.5 nm/V to 0.36 nm/V. This is mostly caused by the reduction.
Fig. 19. Simulated pressure distribution based on the measured PMUT parameters (a) without focus control, (b) focused at 1 mm axial distance, and (c) focused at 1.5 mm axial distance.

of the Q from 167 to 45 due to the additional Parylene layer. The static displacement sensitivity in air, which is approximately the same as the fluid-immersed dynamic displacement (fluid-immersed Q ∼ 1), is only reduced by approximately 50%, as FEM simulation results predict. Reducing the Parylene thickness is expected to improve performance while still sealing the etch holes.

Impedance measurements were conducted using a wafer probe station, transimpedance amplifier (TIA), and network analyzer. Due to the relatively large cable capacitance in the measurement setup, it was difficult to observe resonance-antiresonance frequencies for high frequency (25 MHz) and small PMUTs (25 μm diameter) using this setup. Instead, the electrical impedance of a 1 × 12 linear array of 35 μm diameter PMUTs without Parylene sealing is shown in Figure 16. The amplitude and phase peaks show the resonance of the PMUT membrane, and the resonant frequency matches well with that obtained via LDV measurement (Figure 16).

Figure 17 shows pulse responses of displacement measurements of the 25 μm diameter PMUT (750 nm AlN/800 nm SiO2), driven by a 4 cycles high voltage pulse signal in air (a) and fluid (b). Fluid-immersed measurements were conducted in Fluorinert-70, which has acoustic impedance similar to that of human tissue and high electrical resistivity, eliminating the need for full insulation of all electrical connections to the device. The ring-down time in air and fluid shows the PMUT has a much lower quality factor (Q) in fluid than in air, which is caused by greater damping in fluid. These results match well with the frequency response in fluid, as shown Figure 18. The fluid-immersed transducer has a high 18.6 MHz center frequency and wide 4.9 MHz bandwidth. The center frequency is shifted from 25 MHz in air to 18.6 MHz in fluid because of the increased damping provided by the fluid. This bandwidth is smaller than that of some CMUTs [29], [30] due to the PMUT’s thick membrane (∼1.9 μm, compared to 0.88 μm for the CMUT [29]) and therefore large effective mass. Reducing the PMUT membrane thickness is expected to increase the bandwidth. The peak PMUT membrane vibration velocity is 1.5 mm/s/V, which corresponds to a surface pressure of 2 kPa/V. The output pressure is ∼10× lower than a CMUT with a 120 nm capacitive gap and 40 V DC bias [29]. However, the PMUT doesn’t require a small gap and operation at a high DC bias. Furthermore, the PMUT’s output pressure can be increased by using a thinner Parylene sealing layer, a thinner PMUT membrane or using PZT instead of AlN film.

Phased array simulations were conducted by using Eq. (28) to calculate the pressure field from a single PMUT based on the experimentally measured PMUT velocity. Figure 19 (a) is the pressure distribution without focus control and Figure 19 (b) and (c) are the pressure distributions with the focus set to 1 mm and 1.5 mm focal depth, respectively. This simulation demonstrates the ability to vary the focal point by controlling the phase of the signal applied to the 8 annular rings. Furthermore, it shows a small focus width of 100–150 μm, with acoustic pressures of 9 kPa/V and 6 kPa/V at focus depths of 1 mm and 1.5 mm, respectively. These focus points are ideal for the targeted IVUS imaging application.

VI. CONCLUSION

A 1.2 mm diameter, high fill-factor array of 1,261 AlN PMUTs was fabricated and characterized. At 1061 transducers/mm², the array has a 10–20× higher density than the best PMUT arrays realized to date. FEM and analytical results including resonant frequency, pressure sensitivity, output acoustic pressure and directivity were shown to match well with measurement results. The PMUTs in the
array have center frequencies which match within 0.2 MHz (0.8% of the 25 MHz center frequency), demonstrating good fabrication uniformity. The peak PMUT membrane vibration velocity in the fluid is 1.5 mm/s/V, which corresponds to 2 kPa/V pressure sensitivity. Phased array simulations based on measured PMUT parameters show high output pressure (9 kPa/V) of the focused acoustic beam, demonstrating the feasibility of the array for use in IVUS applications.

ACKNOWLEDGMENT

The authors thank Dr. Benjamin A. Griffin and Keith Ortiz at Sandia National Labs for AlN MEMS fabrication and the UC Berkeley Marvell Nanofabrication Laboratory for post-processing.

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