# HIGH FREQUENCY PIEZOELECTRIC MICROMACHINED ULTRASONIC TRANSDUCER ARRAY FOR INTRAVASCULAR ULTRASOUND IMAGING

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## ABSTRACT

This paper presents a 1.2 mm diameter high fill-factor array of 1,261 piezoelectric micromachined ultrasonic transducers (PMUTs) operating at 18.6 MHz for intravascular ultrasound (IVUS) imaging and other medical imaging applications. At 1061 transducers/mm$^2$, the PMUT array has a 10-20× higher density than the best PMUT arrays realized to date. The PMUTs utilize a piezoelectric material, AlN, which is compatible with CMOS processes. Measurements show a large voltage response of 2.5 nm/V and good frequency matching in air, a 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.

## INTRODUCTION

Ultrasonic transducers are used in many applications, such as nondestructive testing, object detection, gesture recognition and real time medical imaging. Compared with conventional bulk piezoelectric ultrasonic transducers, micromachined ultrasonic transducers (MUTs) have a compliant membrane structure with low acoustic impedance, which is easy to match with the impedance of air and fluids and thereby generate good acoustic coupling. Compared with well-developed capacitive MUTs (CMUTs) [2-3], piezoelectric MUTs (PMUTs) do not require a polarization voltage (which can exceed 190V for CMUTs) 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. 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) films, because of its high piezoelectric coefficient. While PZT has higher piezoelectric constants than the aluminum nitride (AlN) film used in this work, the lower dielectric constant of AlN allows for comparable performance to be achieved, especially in terms of receiver sensitivity [1]. In addition, unlike PZT films that require high fabrication temperature (around 800 $^{o}$C), AlN is deposited at a low temperature (<400 $^{o}$C) and is a material with full compatibility with CMOS processes.

Micromachined transducers 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 through-wafer etching [5-6], resulting in a low fill-factor, small element count, and therefore poor acoustic efficiency. Here we present micromachined transducers with 10-20× higher density (1061 transducers/mm$^2$) than the best PMUT arrays realized to date, 56 transducers/mm$^2$ [5] and 123 transducers/mm$^2$ [6]. This result was achieved using an AlN fabrication process originally developed for RF MEMS resonators, filters [7] and inertial sensors [8]. This process incorporates a sacrificial polysilicon release pit that precisely defines the PMUT diameter, thereby enabling 10× smaller device spacing and eliminating the need for through-wafer etching.

## DESIGN

A PMUT cross-section with exaggerated deformation is shown in Figure 1. As a transmitter, the electric field between the top electrode (TE) and the bottom electrode (BE) creates a transverse stress in the AlN 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 environment. In the receive mode, an incident pressure wave deflecting the plate creates transverse stress which results in a charge between the electrodes due to direct piezoelectric effect. Here, we use AlN as the active piezoelectric layer, which has a low dielectric constant and therefore higher receiving voltage sensitivity. $SiO_2$ is selected to be the passive layer and the layer stack (0.75 µm AlN/ 0.8 µm $SiO_2$) is chosen based on the analysis of neutral axis location and is optimized for high pressure sensitivity. Due to the thin layer stack, a small PMUT with 25 µm diameter has a center frequency of 25 MHz in air and 20 MHz in water. Because the PMUT radius is smaller than λ/2 (λ = 75 µm at 20 MHz in water), the design has the desirable advantage that there is no irregular near-field pressure pattern.

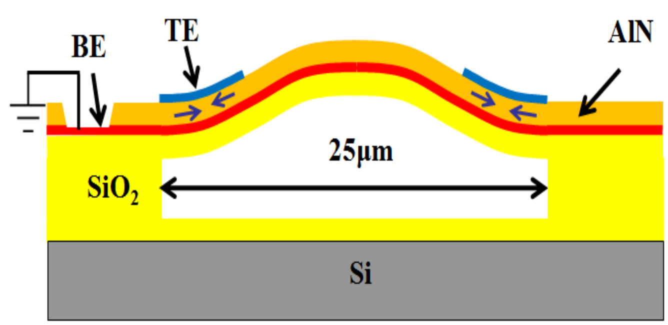


*Figure 1: PMUT cross-section with exaggerated deformation. The PMUT membrane is composed of 750 nm piezoelectric AlN on top of a 800 nm $SiO_2$ passive layer.*

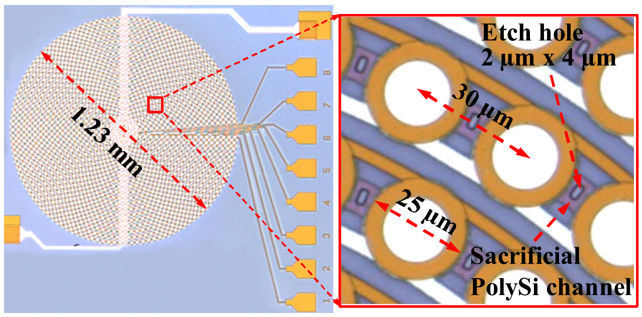


*Figure 2: Optical image of the PMUT array and close-up picture of individual 25 µm diameter PMUTs. The sacrificial poly-Si is removed via 2 µm by 4 µm etch holes between each PMUT.*

The optical image of the fabricated PMUT array and a close-up picture of individual PMUTs are shown in Figure 2. The PMUTs in the 1.2 mm diameter array are grouped into 8 annular rings to enable electronic control of the array's focus by phased array methods. Each group has approximately the same area and number of PMUTs to enable efficient focus depth control. The device is released using etch holes from the front side of the wafer, resulting in a 2 µm cavity beneath the PMUT membrane, similar to the structure used in a CMUT. However, to achieve the same electromechanical coupling ($k_t^2$) in a CMUT, it would require a 6 kV polarization voltage at this 2 µm gap or 50 V with a ~180 nm vacuum gap that is much more difficult to mass produce.

The small 5 µm spacing between PMUTs enables high fill factor. As a result, the array gain, which is the ratio of the array's output pressure to that of a single PMUT, is 20x higher than previously fabricated transducers with the same 25 µm diameter [4]. In addition, compared with conventional ultrasonic transducers using an acoustic lens to obtain a fixed focus, our PMUT array is capable of adjusting the focus depth electronically which improves the imaging resolution across a broad range of medical applications. The small array can be used for high resolution medical imaging, especially intravascular ultrasound (IVUS) imaging, which requires a small size (< 2 mm) of the transducer and high operating frequency (>10 MHz). In IVUS, close proximity to the target allows this PMUT array to pick up a relatively weak ultrasound signal to provide diagnostic information inaccessible from a noninvasive transducer.

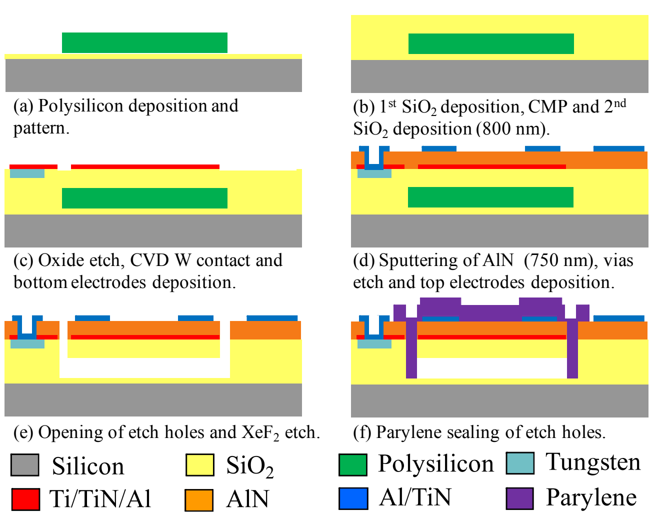


*Figure 3: Fabrication process flow.*

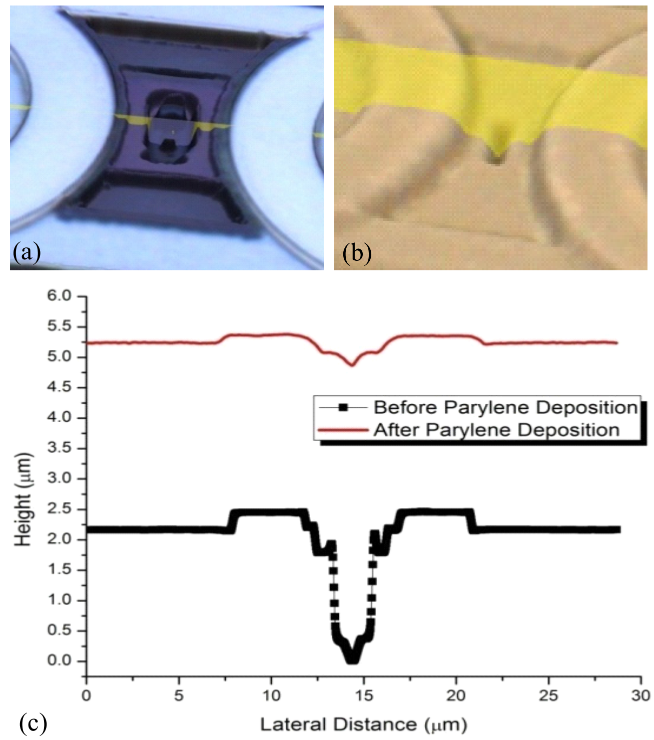


*Figure 4: 3D confocal laser microscope measurements showing a 2 µm by 4 µm etch hole (a) before and (b) after Parylene sealing and (c) profile measurements.*

## FABRICATION

The process flow is shown in Figure 3, where steps (a-d) were performed in the Sandia National Labs AlN MEMS fabrication process 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 µm and even smaller) with close spacing (5 µm) and eliminating the need for through-wafer etching. The sacrificial polysilicon is etched by vapor phase $XeF_2$, releasing the PMUT membranes as shown in step (e), after which the etch holes are sealed and the device is insulated via vapor-phase deposition of Parylene-C to enable fluid immersion, as shown in step (f). The etch holes must be small to enable them to be sealed without filling the cavity beneath the PMUT membrane. Dense arrays having etch holes as small as 2 µm x 4 µm were successfully released. Reducing the density to 155 transducers/mm$^2$ allowed successful release using smaller 1 µm diameter etch holes, suggesting that $XeF_2$ depletion occurs during the release of dense arrays. 3D confocal laser microscope images of an etch hole before and after Parylene sealing are shown in Figure 4, demonstrating that the 3 µm Parylene layer successfully seals the etch hole.

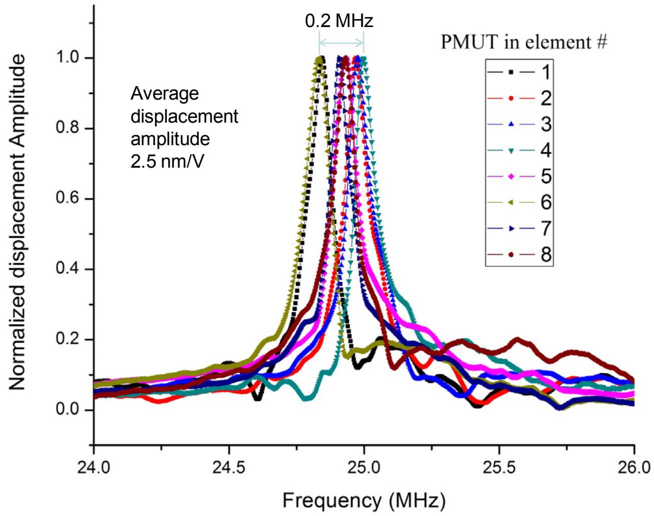


*Figure 5: Frequency response measured in air showing 0.8% frequency mismatch across the PMUT array.*

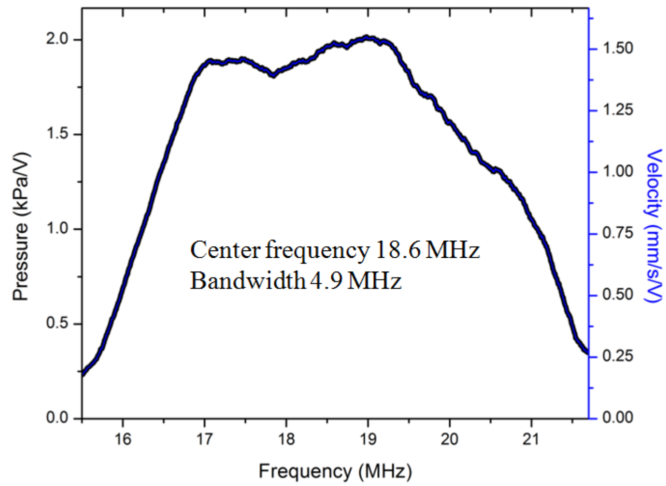


*Figure 7: Measured frequency response in fluid (Fluorinert-70) with velocity converted to pressure on the surface of PMUTs.*

# EXPERIMENTS

A Laser Dropper Vibrometer (OFV 512 and OFV 2700, Polytech) is used in conjunction with a network analyzer (E5061B, Agilent Technologies) to measure the displacement frequency response in air, as shown in Figure 5. The 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 across the array, demonstrating good fabrication uniformity. Measured frequency responses of a single PMUT in air (before and after Parylene sealing) are shown in Figure 6. Although the measurements show that the Parylene layer reduces the dynamic displacement sensitivity from 2.5 nm/V to 0.36 nm/V, it is mostly caused by the reduction of the quality factor (Q) from 167 to 45 due to the additional Parylene layer. The static displacement sensitivity, which is related to the fluid-immersed performance, is only reduced by approximately 50%. Reducing the Parylene thickness is expected to improve performance while still sealing the etch holes.

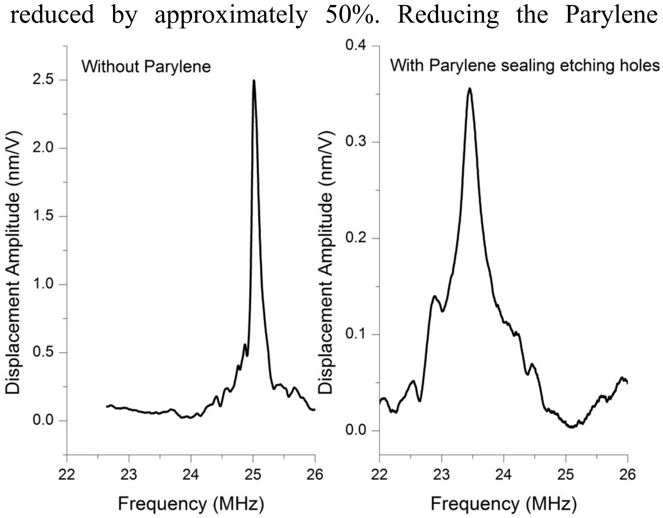


*Figure 6: Frequency response of a single PMUT measured in air before and after Parylene sealing.*

The frequency response in fluid is shown Figure 7. Here PMUTs are immersed in Fluorinert-70, which has similar acoustic impedance with human tissue and a high electrical resistivity, eliminating the need for full insulation of all electrical connections to the device. 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. The wide bandwidth indicates that more energy will be transmitted into the acoustic domain and is indicative of good acoustic coupling. The 0.2 MHz variation of the center frequency across the array is a small fraction of the 4.9 MHz bandwidth and does not greatly affect the array performance. The peak PMUT membrane vibration velocity is 1.5 mm/s/V, which corresponds to a pressure of 2 kPa/V.

Phased array simulations based on the measured PMUT parameters are shown in Figure 8, where Figure 8 (a) is the pressure distribution without focus control and Figure 8 (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 8 annular rings. Furthermore it shows a small focus size of 100 - 150 µm, with acoustic pressures of 9 kPa/V and 6 kPa/V at the focus depths of 1 mm and 1.5 mm respectively. These focus points are ideal for the targeted IVUS imaging application.

# CONCLUSION

A 1.2 mm diameter high fill-factor array of 1,261 CMOS compatible AlN PMUTs is fabricated and characterized. At 1061 transducers/mm$^2$, the PMUT array has a 10-20× higher density than the best PMUT arrays realized to date. The frequency response in air show a peak displacement sensitivity of 2.5 nm/V at the 25 MHz center frequency and

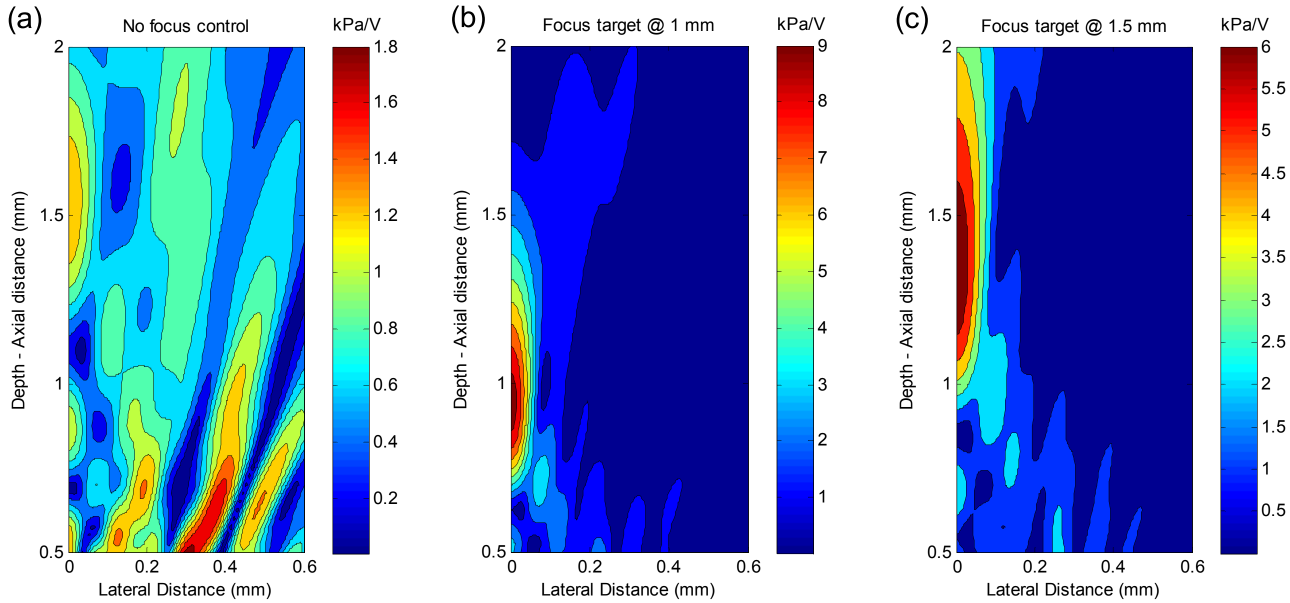


*Figure 8: Simulated pressure distribution based on the measured PMUT parameters (a) without focus control, (b) focused at 1 mm, and (c) focused at 1.5 mm.*

good frequency matching in air 0.2 MHz, demonstrating good fabrication uniformity. The measurements show that the Parylene layer reduces the dynamic displacement sensitivity from 2.5 nm/V to 0.36 nm/V, but it is mostly caused by the reduced quality factor (Q) from 167 to 45 due to the additional Parylene layer. However, the static displacement sensitivity, which is related to the fluid-immersed performance, is only reduced by approximately 50%. Reducing the Parylene thickness is expected to improve performance while still sealing the etch holes. Fluid (Fluorinert-70) immersed measurements reveal a high 18.6 MHz center frequency and wide 4.9 MHz bandwidth, indicating good acoustic coupling. The peak PMUT membrane vibration velocity in the fluid is 1.5 mm/s/V, which corresponds to 2 kPa/V. Phased array simulations based on measured PMUT parameters show high output pressure of the focused narrow acoustic beam, which demonstrates the feasibility of the array for use in IVUS application.


## ACKNOWLEDGEMENTS

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