Capacitive Micromachined Ultrasonic Transducers for Therapeutic Ultrasound Applications

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Abstract—Therapeutic ultrasound guided by MRI is a noninvasive treatment that potentially reduces mortality, lowers medical costs, and widens accessibility of treatments for patients. Recent developments in the design and fabrication of capacitive micromachined ultrasonic transducers (CMUTs) have made them competitive with piezoelectric transducers for use in therapeutic ultrasound applications. In this paper, we present the first designs and prototypes of an eight-element, concentric-ring, CMUT array to treat upper abdominal cancers. This array was simulated and designed to focus 30–50 mm into tissue, and ablate a 2- to 3-cm-diameter tumor within 1 h. Assuming a surface acoustic output pressure of 1 MPa peak-to-peak (8.5 W/cm²) at 2.5 MHz, we simulated an array that produced a focal intensity of 680 W/cm² when focusing to 35 mm. CMUT cells were then designed to meet these frequency and surface acoustic intensity specifications. These cell designs were fabricated as 2.5 mm × 2.5 mm test transducers and used to verify our models. The test transducers were shown to operate at 2.5 MHz with an output pressure of 1.4 MPa peak-to-peak (16.3 W/cm²). With this CMUT cell design, we fabricated a full eight-element array. Due to yield issues, we only developed electronics to focus the four center elements of the array. The beam profile of the measured array deviated from the simulated one because of the crosstalk effects; the beamwidth matched within 10% and side lobes increased by two times, which caused the measured gain to be 16.6 compared to 27.4.

Index Terms—Capacitive micromachined ultrasonic transducers (CMUTs), high-intensity focused ultrasound (HIFU), therapeutic ultrasound, wafer bonding.

I. INTRODUCTION

Therapeutic ultrasound treatment guided by MRI has become popular over the last decade. High-intensity focused ultrasound (HIFU) potentially provides noninvasive treatments that reduce patient morbidity and mortality, lower costs, and widen treatment accessibility [1], [2]. For instance, resection of metastatic colorectal cancer increases 5-year survival from 8% to 30%. However, only 20% of patients are eligible for traditional surgeries. For the remaining 80%, noninvasive treatments could potentially improve outcomes [3], [4].

Compared to thermal ablation tools such as RF, microwave, and laser, ultrasound can be completely noninvasive. Acoustic waves focused into the body can treat a region of interest without harming intervening tissue [5], [6]. Intensities of 500–1000 W/cm² are typical for extracorporeal, focused ultrasound applications for treatment of cancers, and are achieved by designing arrays with gains from 10–100 [6]. MRI is gaining popularity for noninvasive monitoring because it can provide high-resolution anatomical images, accurate thermal temperature maps for real-time monitoring [7], and postoperative functional analysis with diffusion-weighted imaging and contrast-enhanced MRI [8], [9]. A number of MR-guided focused ultrasound systems (MRgFUSs) have been demonstrated for ablation of uterine fibroids, liver cancer, brain tumors, and bone ailments [10]–[12].

Since the 1950s, research groups have developed piezoelectric materials for high-power applications [13], [14]. Using these materials, various transducer configurations have been developed, including mechanically focused transducers [15], unfocused single-element transducers [16], [17], and electronically focused linear and ring arrays [18], [19]. Though piezoelectric transducers are the dominant technology for therapeutic ultrasound, recent advances in the fabrication and design of capacitive micromachined ultrasonic transducers (CMUTs) have made them highly competitive with regard to performance and fabrication flexibility [20], [21]. While many groups have demonstrated CMUTs for imaging purposes [21]–[24], we have been the first group to apply this technology and investigate its advantages for therapeutic ultrasound.

CMUTs are micromachined transducers that are formed from numerous unit cells electrically connected in parallel. Each CMUT cell is a resonator with a moving membrane suspended over a cavity. The cell size and shape [27], [28], membrane topography [29]–[32], and configuration of insulation layers [33], [34] can be modified to optimize performance parameters, such as output pressure, frequency response, and reliability. CMUTs are operated with a dc bias voltage (Vdc) that causes the membranes to deflect downward to a static operating point. This static operating point determines the sensitivity, frequency response, and total acoustic output pressure. In transmit mode, an additional ac voltage is applied to the membrane, which causes the membrane to vibrate around the static operating point and launches pressure waves into the medium.

CMUTs are advantageous for ultrasound applications because the mechanical impedance of the membrane is smaller than the acoustic impedance of water. Because of this, CMUTs are overdamped systems with wide bandwidth and effective
transmission into water [35]. Since therapeutic ultrasound applications frequently excite transducers at a single frequency, a wide bandwidth is not critical, but could be beneficial for applications such as dual-mode ultrasound, where the same transducer is used for both imaging and ablation [36]. In addition, the output pressure of CMUTs is competitive to piezoelectric transducers. Surface acoustic output pressures as high as 1.7–2 MPa peak-to-peak (25–33 W/cm²) at 2.5 MHz have been measured from CMUT devices [37].

CMUTs also exhibit less self-heating than piezoelectric transducers [38], [39]. Piezoelectric transducers have high dielectric losses compared to CMUTs [39], [40]. In addition, unless designed with proper backing and matching layers, acoustic power is lost through the backside in the piezoelectric transducer, whereas most of the power of the CMUT is directed forward [39]. Finally, CMUTs can effectively dissipate small thermal losses because of the high thermal conductivity of silicon (130 W/m K [41]). Piezoelectric materials, such as PZT 4, have a thermal conductivity that is 100 times smaller (1.25 W/m K [44]), and thus, tend to retain heat [39], [41]. For this reason, circulating fluids are often used to cool the piezoelectric transducer [42], [43].

CMUTs are also composed entirely of Type II MR-compatible materials that have magnetic susceptibilities close to water [45]. Previously, we demonstrated that 2.5 mm × 2.5 mm test transducers have a minimal MR artifact, which extended less than 1 mm beyond the edge of the transducer [46]. Minimal artifacts and MR compatibility make CMUTs suitable for use with MR guidance.

Finally, CMUTs are fabricated using silicon micromachining methods with submicrometer accuracy and uniformity, which allow fabrication flexibility. Transducers from 100-μm-sized arrays for intravascular applications to 3-cm-sized arrays for noninvasive therapy with frequencies between 10 kHz–60 MHz have been demonstrated [47]. Since silicon micromachining determines the shapes, sizes, and spacing between neighboring elements, flexible shapes and multiple elements with transducer spacing as small as 3 μm can be fabricated. Silicon micromachining also allows integration with electronics, either monolithically [48] or through chip bonding techniques [49]. Integration of switching and multiplexing electronics simplifies front-end electronics, and reduces parasitic resistances and capacitances associated with cable connections.

Because of these advantages, we designed and developed a CMUT array for therapeutic treatment of upper abdominal cancers. In this paper, we present the simulations and design of an eight-element, equal-area, concentric ring array with a gain of 80. We also present the simulation and design of a circular CMUT cell with thick conductive membranes capable of surface acoustic output pressures up to 1.4 MPa peak-to-peak (16.3 W/cm²) at 2.5 MHz. With this gain and surface acoustic output power, we can hope to achieve the 500–1000 W/cm² yield as needed for therapeutic ultrasound [6]. Based on these designs, test transducers and a concentric ring array were fabricated. The test transducers were used to verify the CMUT cell models and design before testing the full concentric ring array. After fabrication of the eight-element ring array, we found that the yield was lacking, but we were able to demonstrate continuous wave (CW) focusing with the four center elements. A gain of 16.6 and a maximum focal intensity of 85 W/cm² was achieved with only four out of eight elements, which matches well with the acoustic models.

II. METHODS

An eight-element, equal-area concentric ring array design (see Fig. 1) was chosen for a first prototype for upper abdominal cancer ablation [50]. Though a complete 2-D array is ideal for flexibility in focusing and steering the ultrasound beam, this design was chosen to reduce the number of drive channels, while maintaining focal intensities near 500–1000 W/cm² needed for ablation [6]. The array was designed to focus 30–50 mm inside the body and ablate a 4-mm-diameter lesion in the focal plane in 30 sec. By moving the focus of the transducer, lesions can be tiled to treat the entire tumor volume. To treat a 2- to 3-cm-diameter cancer tumor with this lesion size requires roughly 1 h [50]. This concentric ring design was first developed and refined through simulation to determine the output pressure and frequency required from the individual CMUT cells. These cells were then designed and simulated using a finite-element model (FEM). Test transducers (2.5 mm × 2.5 mm) were used to confirm these models and the CMUT cell performance; afterwards, eight-element arrays were fabricated and tested. Because of issues with yield, only four out of the eight elements were used for testing and measurements.

A. Simulation

1) Array Simulation: MATLAB (Mathworks, Matick, MA) was used to calculate the beam profile of the array in homogenized liver (see Table I [21], [51]) using Huygen’s Principle, which estimates each array element as a collection of point sources [52], [53]. The pressure due to each point source is given by

\[
p_i(x, y, z) = \sqrt{\frac{2W \rho fS}{cA d}} e^{-(\phi-(\phi/2\pi d/\lambda_t+\phi/2))}
\]

(1)

Fig. 1. Photograph of the eight-element concentric ring array. The array has front-side traces and bonding pads.
where $W$ is the total acoustic power from the array, $A$ is the area of the array, $S$ is the area of each source, $\rho$ is the density of the medium, $c$ is the speed of sound in the medium, $f$ is the frequency, $\lambda$ is the wavelength, and $d$ is the distance from the simple source to the point of interest [53]. The variable $i$ is equivalent to $\sqrt{-1}$. The phase of each element was calculated by accounting for the difference in propagation time between the centers of each element and is given by [54]

$$\phi_n = \frac{c (x_n - z)}{c}$$

where $z$ is the focal length, $x_n$ is the distance to the center of the $n$th ring, $c$ is the speed of sound in the medium, and $\omega$ is the frequency of operation. The total pressure at each point was calculated by dividing the transducer area into a simple sources spaced $\lambda/10$ apart and summing the response from every source to the point of interest to produce a pressure profile as

$$P(x, y, z) = \Sigma P_i(x, y, z).$$

To calculate the temperature and heating due to the pressure profiles, the intensity $I$ was calculated at each point by the following equation [6]:

$$I = \frac{P_{peak}^2}{2Z}$$

where $Z$ is the impedance of the medium and $P_{peak}$ is the peak pressure at the points of interest, assuming a sinusoidally varying pressure. Intensities were converted to power per unit volume using the following equation, where $\alpha$ is the absorption of tissue, taken to be approximately one-third of the attenuation [55]

$$q_m = 2\alpha I.$$  (5)

The Pennes Bioheat equation [56] was used to determine the temperature in a homogeneous region of liver tissue over time using the power per unit volume of the pressure field

$$\rho c_l \frac{\partial T}{\partial t} = \nabla (k_l \nabla T) + c_b w_b (T_a - T) + q_m$$  (6)

where $\rho$ is the density of liver, $c_l$ is the heat capacity of liver, $k_l$ is the conduction coefficient of liver, $c_b$ is the heat capacity of blood, $w_b$ is the perfusion coefficient, $T_a$ is the body's ambient temperature (37°C), and $T$ is the temperature as a function of time. The values of these parameters are given in Table I. This partial differential equation was solved by calculating the second derivative of temperature with regard to space at each time point. Then, at each point in space, the resulting ordinary differential equation can be solved to reach the successive time point.

After calculating the temperature, tissue necrosis was determined by using the cumulative equivalent minutes at 43 °C (CEM$_{43}$) [57]. This metric determines tissue death by estimating total temperature dose over time and is given by

$$CEM_{43} = \sum_{t=0}^{t_{\text{final}}} \left( R(43-T(t)) \right) \Delta t$$  (7)

where $T(t)$ is the the temperature over time. $R = 0.25$ if $T(t) < 43$ and $R = 0.5$ if $T(t) > 43$. To necrose liver tissue, a CEM$_{43}$ threshold of 240 equivalent minutes is needed [6].

Using an acoustic surface pressure of 1 MPa peak-to-peak (8.5 W/cm$^2$) at 2.5 MHz, we calculated the beam profile of the 30-mm-diameter, eight-element concentric ring array. The array was flat, and the inner and outer radius of each element was chosen so that every element had the same area, 88 mm$^2$.

2) CMUT Cell Simulation: Based on the frequency (2.5 MHz) and surface acoustic pressure [1 MPa peak-to-peak, 8.5 W/cm$^2$]) determined from the acoustic wave simulation of the array, the CMUT cells were designed and simulated using FEM. We used a cylindrical waveguide, 2-D axisymmetric FEM to simulate the cell [58] (see Fig. 2). We used predefined element types from ANSYS (ANSYS 8.0, Canonsburg, PA) to develop the FEM. PLANE42 elements [59] were used to model the CMUT cell, while FLUID29 elements were used to model a nonattenuating, nonabsorbing fluid column placed on the surface of the cell. TRANS126 elements are electromechanical elements used to convert electrical inputs to mechanical force. The model was meshed so that the radius of the cell was partitioned into 32 nodes and each element was no larger than 2 $\mu$m in size.

The boundary conditions were set such that the nodes along the bottom of the cell were stationary in $y$, while the nodes along

| TABLE 1 |
|-----------------|-----------------|-----------------|
| speed of sound (m/s) | soybean oil | homogenized liver | blood |
| density (kg/m$^3$) | 1430 | 1550 | - |
| attenuation (dB/cm) | 930 | 1060 | 1000 |
| heat capacity (J/kg-K) | 0.47 | 0.98 | 1.85 |
| conductivity (W/K-m) | - | 3600 | 3840 |
| blood perfusion rate (kg/m$^3$/s) | 0.5 | - | 0.5 |
the outer boundary were immovable in $x$. An absorbing boundary was placed at the end of the fluid column, three wavelengths away from the surface of the CMUT. By using these boundary conditions, we assume that an infinite number of cells surround the modeled cell, which makes use of the periodic structure of the CMUT. This waveguide model assumes that all the cells are exactly in phase and that modes higher than the cutoff frequency of the circular waveguide’s first mode, the $(0, 1)$ mode, are not present [60]. When these higher order modes exist, the conditions of the absorbing boundary become invalid [58].

In the model, the static dc voltage was first applied, and the resulting displacement and stresses were calculated. To determine the frequency response of the device, a prestressed harmonic analysis was used, in which a small signal voltage with varying frequencies was applied [59]. The output pressure was averaged across nodes that were half a wavelength from the surface of the cell and plotted against frequency. For a transient analysis, large signal sinusoidal voltages with five cycles were applied across nodes that were half a wavelength from the surface of the CMUT cell. The average pressure was also calculated by averaging the pressure from each node half a wavelength from the surface of the cell and plotted as a function of time.

\[ \text{B. Fabrication} \]

We used a direct wafer bonding process that combines bulk and surface micromachining [20] to fabricate the therapeutic devices. The simplest wafer bonding process (see Fig. 3) begins with two wafers, a prime quality silicon wafer and a silicon-oxide (SOI) wafer, both heavily doped for electrical conductivity. Doping the SOI wafer allows the silicon layer to be used as the top electrode, thereby eliminating the metal from the membrane surface. This eliminates problems associated with electromigration. First, the cavity is defined on the prime wafer using thermal oxidation and photolithographic patterning of the oxide layer. A second oxide layer is thermally grown to form the insulation layer, which protects the CMUT against electrical breakdown. Since oxidation of a nonplanar surface produces bulges at the edge of the cavities, a second etch is needed to remove the edges and finalize the cell size. Because of the nonlinearity in the rate of thermal oxidation, gaps with high precision can be realized with these two thermal oxidation steps. Afterwards, the wafer is fusion-bonded to an SOI wafer. The handle and the buried oxide (BOX) layer of the SOI wafer are removed by grinding and wet etching with tetramethyl ammonium hydroxide, leaving a single-crystal silicon membrane [20]. Afterwards, ground contacts are etched, and aluminum electrodes are deposited.

Test transducers and eight-element concentric ring arrays were fabricated with a close-packed arrangement of a circular cell with 70 $\mu$m radius, 6 $\mu$m conductive membranes, and 0.4 $\mu$m large gaps. After fabrication, the devices were diced and wire-bonded to a printed circuit board (PCB) for characterization and testing. The wire bonds were covered with 5-min epoxy to protect them mechanically. The test transducers were 2.5 mm by 2.5 mm square and were used to characterize the output pressure and frequency response of the CMUT cell design. After verifying our CMUT cell model, an eight-element, equal-area concentric ring array with 30 mm diameter was fabricated and operated to demonstrate focusing.

\[ \text{C. Measurement} \]

1) \textbf{Test Transducer Measurements:} Frequency response and dynamic response of the test transducers were measured to verify the cell design and model. For these measurements, transducers were immersed in soybean oil (attenuation of 0.438 dB/cm at 2.5 MHz [21], [25] and speed of sound of 1430 m/s [26]). Soybean oil provides electrical insulation and has similar properties to liver tissue [21]. The glass container used to hold the CMUT and oil was larger than the device to reduce reflections from the container walls and oil–air interface.

Frequency response and dynamic response were measured using a hydrophone setup (see Fig. 4). A HNP-0400 hydrophone
For HIFU operation, using an air core inductor for efficient power transfer. \( \Omega \) with an series 2 stage. For beam profile measurements, the electronics developed to drive from each element to produce constructive interfere of the wave CW focusing involves shifting the phase of the pressure wave at 2.5 MHz was applied to the transducer for periods of at tions. A dc voltage of 172 V and a 250-Vpp sinusoidal signal applied to each transducer element, which is matched to 50 \( \Omega \) with an series inductor. While the system is easily expandable to eight channels, since only four elements were active, we only show the electronics connections for four channels in this figure.

(Onda Corporation, Sunnyvale, CA) was positioned 2 cm from the surface of the transducer to measure the output pressure in the far field. The frequency response of the transducer was determined by applying a single 10-V peak-to-peak (Vpp), 20 MHz, bipolar, square-wave input and a dc bias voltage that was 172 V (70% of the pull-in voltage). This input approximated an impulse. After applying a fast Fourier transform to the measured signal and correcting the result with the frequency response of the hydrophone [63], we determined the transducer’s frequency response. This measurement was then compared to the prestressed harmonic response calculated from the FEM.

For the dynamic response, both burst measurements and CW were performed. Burst measurements were made with an input signal that was a sinusoidal, 30-cycle signal between 25–300 Vpp at 2.5 MHz. We used a hydrophone to measure the response 2 cm from the surface of the transducer. This measurement was then used to calculate the output pressure at the surface of the transducer by accounting for diffraction and attenuation through the soybean oil [62], as well as the transfer function of the hydrophone [63]. These measurements were plotted versus applied ac voltage and compared with the calculated transient response. CW measurements were made to test the ability of the device to operate for CW therapeutic ultrasound applications. A dc voltage of 172 V and a 250-Vpp sinusoidal signal at 2.5 MHz was applied to the transducer for periods of at least 1 h.

2) Transducer Array Measurements: For HIFU operation, CW focusing involves shifting the phase of the pressure wave from each element to produce constructive interfere of the wave fronts at the focal point. The electronics developed to drive these elements (see Fig. 5) consisted of a direct digital synthesizer (DDS9m, Novatech Instruments, Inc., Seattle, WA), four-channel phasing board, capable of sinusoidal signals up to 170 MHz with a resolution of 0.1 Hz. The DDS9m also has 14 bit programmable phase, translating to a resolution of 0.022\(^\circ\). The ac signal was then amplified by 1 W (ZHL-3A-S) amplifiers (Minicircuits, Brooklyn, NY); a second bank of custom-designed 20 W amplifiers were also built and used when higher power was needed. The ac signal from the amplifiers was then superimposed with a dc bias voltage using a bias T, consisting of a dc blocking capacitor and resistor. The resulting signal was applied to the CMUT elements, each of which was tuned to 50 \( \Omega \) using an air core inductor for efficient power transfer.

The transducer array was immersed in a tank of soybean oil, which was then placed on a motorized and electronically controlled \( x \)-\( y \) stage. For beam profile measurements, the electronics system was connected to the four center elements and the phases were changed to focus the array to 35 mm. Only the four center elements were used because of yield issues, as explained in the results and discussion sections.

In order to measure the beam profile, a CW, 15 Vpp signal with bias voltage of 140 V was applied to each element. A hydrophone (HNV-0400, Onda Corporation, Sunnyvale, CA) was immersed in oil and positioned 35 mm from the surface of the transducer. The array was scanned in the focal plane over a 15 mm\(^2\) area with 0.2 mm step size surrounding the focal point to form a beam profile in the focal plane. The measured beam profile was compared with a simulated beam profile of the four center elements, which was calculated using Huygen’s principle. The power gain was also calculated by averaging the intensity over the focal region; this was then compared to the model.

III. RESULTS

A. Acoustic Simulation

The beamwidth of the transducer was 1.2 mm in the focal plane and 9 mm in the axial direction (see Fig. 6) when focusing to 35 mm. The power gain of the array is 80 at the focus, which translates to 680 W/cm\(^2\) spatial average over the \(-6\) dB beamwidth with the surface acoustic pressure of 1 MPa peak-to-peak (8.5 W/cm\(^2\)). After calculating the heating profile, we found that 5-mm lesions in liver tissue can be formed within 30 s (see Fig. 7). Lesions can be successively created to treat the total volume of the tumor.

These results set the design requirements for the CMUT devices. To account for cooling in large blood vessels and variability in perfusion, a targeted surface acoustic pressure of 1.4 MPa peak-to-peak (16.3 W/cm\(^2\)) is necessary [50]. The previous simulation assumed that there was no tissue motion. This assumption holds if the application of therapy is gated to the breath-holds of the patient. If no gating is used, the tissue moves cyclically, and the lesion formed is elliptical. The spot size is narrower because the power is spread over a larger area.

B. CMUT Cell Simulation and Testing

Using FEM, a circular cell (see Fig. 8) with 70 \( \mu \)m radius, 6 \( \mu \)m conductive membrane, and 0.4 \( \mu \)m large gaps was designed to operate at 2.5 MHz with an acoustic output pressure up to 1.75 MPa peak-to-peak (25.5 W/cm\(^2\)). Frequency response and transient measurements were made and compared to this simulation. The measured center frequency and bandwidth matched the simulated values within 5.2% and 5.6%, respectively.
Fig. 6. Simulated beam profile in homogenized liver of the eight-element concentric ring array (a) in the focal plane (30 dB dynamic range) and (b) along the axial direction simulated using Huygen’s principle. The beamwidth of the transducer was 1.2 mm in the focal plane and 9 mm in the axial direction. The transducer had a focal gain of 80, which implies that a 1-MPa peak-to-peak surface acoustic pressure (8.5 W/cm²) will produce a focal intensity of 680 W/cm².

Fig. 7. Simulated necrosed area in the focal plane as a function of time shows that a 5-mm-diameter lesion can be formed in 30 s.

(see Fig. 9). Discrepancy between the measurement and simulation could be caused by acoustic crosstalk or inaccuracies of the hydrophone transfer function. In addition, the model that we used assumes an infinite transducer with close-packed cells, which will, in general, make the bandwidth wider than a finite transducer. In order to accurately simulate a finite transducer, however, it would require a 3-D model that is computationally intensive and impractical for designing such transducers.

Fig. 8. Schematic of the circular cells used in our design. (a) Cells were patterned in a close packed arrangement into both test transducers and concentric ring arrays. (b) Cross section of an individual cell shows the basic form of the resonator, a conductive silicon membrane suspended over a vacuum gap formed in oxide.

Fig. 9. Comparison of the measured and simulated frequency response of the circular CMUT design. The center frequency and bandwidth match within 5.2% and 5.5%, respectively. The discrepancy between the measured and simulated profile could be caused by factors not accounted for in the model, such as acoustic crosstalk and also the finite size of the element.

Transient analysis of the surface output pressure with respect to the normalized peak-to-peak ac voltage showed discrepancy between measured and simulation because of two main factors (see Fig. 10). The discrepancy of the output pressure was caused by charging of the silicon dioxide in the gap of the cell. Accumulation of sheet charge in this layer causes the operation point of the device to change. This means that increased voltages have to be applied to achieve the expected pressures [33]. Crosstalk can also reduce the effective output pressure because it causes neighboring cells to operate asynchronously, decreasing the average output pressure [64]–[68].

In spite of the reduction of output pressure from the expected, we fabricated a 2.5-MHz transducer with an output pressure of 1.4 Mpa peak-to-peak (16.3 W/cm²) at a dc voltage of 172 V (70% of the pull-in voltage) and ac peak-to-peak voltage greater...
than 100% of the pull-in voltage. With this surface acoustic pressure and the gain of the array (theoretically 80), we should be able to achieve the 500–1000 W/cm² needed for HIFU. Though the ac voltage exceeds 100% of the pull-in voltage, the average output pressure is still sinusoidal at the fundamental frequency, and also contains higher frequency components (less than 15% for our transducer). The ac signal does not cause the membrane to collide with the bottom of the cavity. The behavior of the membrane at these high frequencies is a function of the frequency of actuation, the membrane mechanical properties, and the dc voltage, which need to be calculated with FEM [69].

We also operated the test transducer in CW mode. When the device was driven with a dc voltage of 196 V (80% of pull-in voltage) and a CW ac voltage of 250 Vpp at 2.5 MHz, the CW surface output pressure was 1.7 MPa peak-to-peak (20 W/cm²) over 1 h [38].

C. Array Testing

Though an eight-element array was fabricated (see Fig. 1), only the four center elements were used because of yield issues from processing in a multiuser fabrication facility. Oxide defects and unbonded areas from particles on the wafer’s surface caused elements to become shorted. While 2.5 mm × 2.5 mm test transducers showed a 97.5% yield, because the array elements had a large area (88 mm²), the same yield on an area basis means that on average, every one to two array element will have at least one small defect.

Because we could only measure the four center elements of the ring array, we resimulated the acoustic profile of the four-element ring array in oil to be able to accurately compare to the measured beam profile. The measured beam profile matches well with the expected profile (see Fig. 11). The lack of symmetry on the right in the beam profile is caused by the asymmetry in the actual array due to the need for front-side interconnects. Looking at a slice of the beam profile through the diameter of the array, one sees that the measured beamwidth is 10% smaller than the simulated one and the sidelobe level is twice as large. From crosstalk measurements, cells in neighboring elements show a center displacement at least 10 dB down from the displacement of the excited element. We found surface waves traveling at 1182 m/s, which are related to dispersive guided modes. This acoustic crosstalk causes the effective aperture to be increased, which slightly decreases the beamwidth, as observed in the beam profile. In addition, acoustic crosstalk waves also cause phase errors in neighboring elements, which increases the sidelobe energy [62]. Because of this, the power gain of the four elements was calculated to be 27.4 and measured to be 16.6. We were able to increase the surface acoustic pressure so that the focal intensity was 85 W/cm² before oxide defects caused a significant amount of elements to break down.

IV. DISCUSSION

The agreement of the array with the acoustic model is promising for developing arrays for therapeutic ultrasound. However, the focal gain of the four center elements, 16.6, was too small to perform HIFU. The focal intensity was so small because half of the area of the array was destroyed by micrometer- to millimeter-sized oxide and bonding defects. Though the yield for 2.5 mm × 2.5 mm devices was 97.5%, on a per area basis, because the array element size is 88 mm², this translates to a defect for roughly every one to two elements. If the array design were modified to reduce the effects of the defects and increase the active area of the device, the focal gain should be improved.
to a gain near 80, as we simulated. Also, we will be able to operate the array with larger voltages and at least three times more surface acoustic output pressures without oxide breakdown and defects. This would enable us to increase the output intensity to levels near 1000 W/cm² needed for HIFU.

The majority of defects are small compared to the area that they cause to fail. The two main causes of defects are oxide defects that cause dielectric breakdown and bonding defects where the membrane peels off the device during processing. Oxide defects are usually small, often several hundred micrometers. This is because they are often caused by particles during oxidation or lattice defects. Bonding defects are larger, on the order of several millimeters, and can be caused by contaminants on the wafer’s surface that prevent the membrane from making contact with the base wafer during the bonding process [20].

Both design and processing modifications can mitigate the effects of defects. With regard to design, arrays with smaller element sizes, on the order of several millimeters, could be constructed. Examples of such arrays include a circular array that is sectored or a full 2-D array. In the array, elements with defects can be identified and disconnected from the electronics. In this way, a small defect would only remove one small element, which has a minimal effect on the beam profile.

Process improvements like hydrophilic surface treatments [70], [71], which improve the bond strength between two wafers, can be used to mitigate bonding problems. To improve the oxide breakdown qualities, a local oxidation process that selectively grows thick oxide insulation layers can be used to increase the oxide thicknesses [34]. This decreases the electric field in the oxide and reduces the chance of oxide breakdown. Finally, the entire device can be passivated with nitride to protect the oxide from being exposed to the moisture and dirt in the environment. Moisture can cause the oxide to degrade and breakdown. With these modifications, a ring or 2-D array for HIFU ablation of upper abdominal cancers can be successfully fabricated.

In future work, the CMUT cell design could potentially be improved to increase surface acoustic output pressure to reach the needed 1000 W/cm². Though conventional CMUT cells were shown to exhibit properties necessary for ultrasound therapy, both the average output pressure and reliability can be improved, using different membrane topologies [29]–[32] and configuration of insulation layers [33], [34], respectively.

While the sidelobe energy of our profile has a maximum of −10 dB, because tissue necrosis is an exponential process, suppression to these levels is adequate. Having a greater number of elements and greater degree of focusing would greatly reduce these sidelobe levels [62]. In addition, creating walls [72] and coating the array with attenuating layers [73] could also reduce the effects of interelement crosstalk and improve the beam profile of our array.

This array was designed to produce 5-mm lesions in 30 s in order to ablate a 2- to 3-cm-diameter tumor in about 1 h. Though this treatment time and method were chosen for this paper, study of treatment time requires in vivo studies, and the determination of the optimal treatment method is beyond the scope of this paper.

V. CONCLUSION

CMUTs are promising for therapeutic ultrasound applications. In this paper, we have demonstrated the simulation, design, and development of an eight-element, equal-area concentric ring array for treatment of upper abdominal cancers. This array was simulated to have a gain of 80 and can form 5-mm lesions in 30 s. CMUT cells with 70 µm radius, 6 µm membrane thickness, and 0.4 µm gap were designed for this array and shown to produce acoustic output pressures of 1.4 MPa peak-to-peak (16.3 W/cm²) at 2.5 MHz. An eight-element concentric ring array with equal area elements was fabricated using this cell design. Though yield issues prevented the full eight elements from functioning, we operated the four center elements with custom designed electronics. The beam profiles of the measured and simulate response were similar. Because of crosstalk, the measured width of our array was 10% smaller than the simulated response, while the sidelobe level was twice as high. The four elements showed an acoustic intensity of 85 W/cm² at the focus, which could be improved if the active area of the device were increased.

With this successful start, sophisticated 2-D arrays with smaller element sizes can be used to reduce the problem of micrometer- to millimeter-sized defects. With smaller elements, the removal of one element due to a defect would not greatly affect the beam profile of the transducer. In addition, movement to a 2-D array with integrated electronics would allow full beam steering and complex beam forming methods that are useful for focusing around ribs and air when targeting applications near the lungs in the liver.

REFERENCES


Serena H. Wong (M’08) was born in Stanford, CA. She received the B.S. and M.S. degrees in electrical engineering in 2002 and 2003, respectively, from Stanford University, Stanford, where she is currently working toward the Ph.D. degree in electrical engineering at the E. L. Ginzton Laboratory.

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