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Capacitive Micromachined Ultrasonic Transducers for Therapeutic Ultrasound Applications

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Abstract

Therapeutic ultrasound guided by magnetic resonance imaging (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 8-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 2–3 cm diameter tumor within 1 hour. 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 by 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 8-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 because of crosstalk effects; the beamwidth matched within 10% and sidelobes increased by 2 times, which caused the measured gain to be 16.6 compared to 27.4.

Index Terms

Therapeutic ultrasound; HIFU; wafer bonding; CMUTs; charging; performance

I. Introduction

Therapeutic ultrasound treatment guided by magnetic resonance imaging (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 radiofrequency (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 post-operative functional analysis with diffusion-weighted imaging and contrast enhanced MRI [8], [9]. A number of MR guided focused ultrasound systems (MRgFUS) have been demonstrated for ablation of uterine fibroids, liver cancer, brain tumors, and bone ailments [10]–[12].

Since the 1950's, 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 (V_{DC}) 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 operation 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 of entirely Type II MR compatible materials that have magnetic susceptibilities close to water [45]. Previously, we demonstrated that 2.5 by 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 makes CMUTs suitable for use with MR guidance.

Finally, CMUTs are fabricated using silicon micromachining methods with sub-micron 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 8-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 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 8-element ring array, we found that the yield was lacking, but we were able to demonstrate continuous wave focusing with the four center elements. A gain of 16.6 and a maximum focal intensity of 85 W/cm² was achieved with only 4 out of 8 elements, which matches well with the acoustic models.

II. Methods

An 8-element, equal-area concentric ring array design (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 seconds. By moving the focus of the transducer, lesions can be tiled to treat the entire tumor volume. To treat a 2–3 cm diameter cancer tumor with this lesion size requires roughly 1 hour [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 by 2.5 mm) were used to confirm these models and the CMUT cell performance; afterwards, 8-element arrays were fabricated and tested. Because of issues with yield, only 4 out of the 8 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 (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

$$p_i(x, y, z) = \sqrt{\frac{2W\rho}{cA}} \frac{fS}{d} e^{((\phi - 2\pi d/\lambda)i - d\alpha)}, \quad (1)$$

where W is the total acoustic power from the array, A is the area of the array, S is the area of each source, ρ is the density of the medium, c is the speed of sound in the medium, f is the frequency, λ is the wavelength, and d is the distance from the simple source to the point of interest [53]. The variable, i , is equivalent to the $\sqrt{-1}$. 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]:

$$\phi_n = \frac{\omega(x_n - z)}{c}, \quad (2)$$

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 ω 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:

$$P(x, y, x) = \sum p_i(x, y, z) \quad (3)$$

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

$$I = \frac{P_{peak}^2}{2Z}, \quad (4)$$

where Z is the impedance of the medium and the 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 α is the absorption of tissue, taken to be approximately one third of the attenuation [55]

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

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.

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

where ρ 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 (CEM₄₃) [57]. This metric determines tissue death by estimating total temperature dose over time and is given by,

$$CEM_{43} = \sum_{t=0}^{t=final} R^{(43-T(t))} \Delta t, \quad (7)$$

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 CEM₄₃ threshold of 240 equivalent minutes is needed [6].

Using an acoustic surface pressure of 1 MPa peak to peak (8.5 W/cm²) at 2.5 MHz, we calculated the beam profile of the 30 mm diameter, 8-element concentric ring array. The array was flat and the inner and outer radius of each element was chosen so every element had the same area, 88 mm².

2) CMUT cell simulation—Based on the frequency (2.5 MHz) and surface acoustic pressure (1 MPa peak to peak, 8.5 W/cm²) 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] (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 μ m in size.

The boundary conditions were set so that the nodes along the bottom of the cell were stationary in y , while the nodes along 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 5 cycles were applied to 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.

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 (Fig. 3) begins with two wafers, a prime quality silicon wafer and a silicon-on-oxide (SOI) wafer, both heavily doped for electrical conductivity. Doping the silicon on insulator (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 8-element concentric ring arrays were fabricated with a close-packed arrangement of a circular cell with 70 μm radius, 6 μm conductive membranes, and 0.4 μm large gaps. After fabrication, the devices were diced and wirebonded to a printed circuit board (PCB) for characterization and testing. The wire bonds were covered with 5-minute 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 8-element, equal-area concentric ring array with 30 mm diameter was fabricated and operated to demonstrate focusing.

C. Measurement

1) 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 (Fig. 4). A HNP-0400 hydrophone (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 then 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 hour.

2) Transducer Array Measurements—For HIFU operation, CW focusing involves shifting the phase of the pressure wave from each element to produce constructive interference of the wave fronts at the focal point. The electronics developed to drive these elements (Fig. 5) consisted of a direct digital synthesizer (DDS9m, Novatech Instruments, Inc., Seattle, WA), 4 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 degrees. 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 ohms 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 (HNP-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 squared 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 (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 -6dB 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 seconds (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, then the tissue moves cyclically and the lesion formed is elliptical and the size of 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 (Fig. 8) with $70 \mu\text{m}$ radius, $6 \mu\text{m}$ conductive membrane, and $0.4 \mu\text{m}$ 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. 9). Discrepancy between the measurement and simulation could be caused by acoustic crosstalk or inaccuracies of the hydrophone transfer function. In addition, the model 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.

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 (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/cm^2) 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/cm^2 needed for HIFU. One can see that when the AC peak to peak voltage exceeds 100% of the pull-in voltage, then these pressures can be achieved. Though the AC voltage exceeds 100% of the pull-in voltage, the average output pressure is still sinusoidal at the fundamental frequency, but 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 V_{pp} at 2.5 MHz, the CW surface output pressure was 1.7 MPa peak to peak (20 W/cm^2) over one hour [38].

C. Array testing

Though an 8-element array was fabricated (Fig. 1), only the four center elements were used because of yield issues from processing in a multi-user fabrication facility. Oxide defects and un-bonded areas from particles on the wafer's surface caused elements to become shorted. While 2.5 mm by 2.5 mm test transducers showed a 97.5% yield, because the array elements had a large area (88 mm^2), 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 4 center elements of the ring array, we re-simulated the acoustic profile of the 4-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 (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 side lobe level is twice as large. From crosstalk measurements, cells in neighboring elements showed 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 decreases the beam width slightly, as observed in the beam profile. In addition, acoustic crosstalk waves also cause phase errors in neighboring elements, which increases the side lobe energy [62]. Because of this, the power gain of the 4 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/cm^2 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 4 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 micron to millimeter sized oxide and bonding defects. Though the yield for 2.5 mm by 2.5 mm devices was 97.5%, on a per area basis, because the array element size is 88 mm^2 , 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/cm^2 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 microns. 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 sectorized 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 1000 W/cm^2 needed. 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 side lobe 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 side lobe levels [62]. In addition, creating walls [72] and coating the array with attenuating layers [73] could also reduce the effects of inter-element crosstalk and improve the beam profile of our array.

This array was designed to produce 5 mm lesions in 30 seconds in order to ablate a 2–3 cm diameter tumor in about one hour. 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 8-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 seconds. 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/cm^2) at 2.5 MHz. An 8-element concentric ring array with equal area elements was fabricated using this cell design. Though yield issues prevented the full 8 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 side lobe level was twice as high. The 4 elements showed an acoustic intensity of 85 W/cm^2 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 micron 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

1. Gillett MD, Gettman MT, Zincke H, Blute ML. Tissue Ablation Technologies for Localized Prostate Cancer. *Mayo Clinical Proceedings*. 2004; 79(no. 12):1547–1555.
2. Archer SB, Sims MM, Giklich R, Traverso E, Laycock B, Wolfe BM, Apfelgren KN, Fitzgibbons RJ, Humer JG. Outcomes assessment and minimally invasive surgery. *Surgery Endoscopy*. 2000; 14:883–890.
3. Butler J, Attiyyeh FF, Daly JM. Hepatic resection for metastases of the colon and rectum. *Surgery Gynecology Obstetrics*. 1986; 162(no. 2):109–113.
4. Hughes KS, Rosenstein RB, Songhorabodi S, Adson MA, Ilstrup DM, Fortner JG, Maclean BJ, Foster JH, Daly JM, Fitzherbert D. Resection of the liver for colorectal carcinoma metastases. A multi-institutional study of long-term survivors. *Dis Colon Rectum*. 1988; 31(no. 1):1–4. [PubMed: 3366020]
5. Kennedy JE, ter Haar GR, Cranston D. High Intensity Focused Ultrasound: Surgery of the Future? *British Journal of Radiology*. 2003; 76:590–599. [PubMed: 14500272]
6. ter Haar G. Ultrasound Focal Beam Surgery. *Ultrasound in Medicine and Biology*. 1995; 25(no. 5): 1089–1110. [PubMed: 8849823]
7. Rieke V, Butts Pauly K. MR Thermometry. *Journal of Magnetic Resonance Imaging*. 2008; 27(no. 2):376–290. [PubMed: 18219673]

8. Cline HE, Schenck JF, Hynynen K, Watkins RD, Souza SP, Jolesz FA. MR-guided focused ultrasound surgery. *Journal of Computer Assisted Tomography*. 1992; 16(no. 6):956–965. [PubMed: 1430448]
9. Cline HE, Hynynen K, Hardy CJ, Watkins RD, Schenck JF, Jolesz JF. MR Temperature Mapping of Focused Ultrasound Surgery. *Magnetic Resonance in Medicine*. 1994; 31(no. 6):628–636. [PubMed: 8057815]
10. Hesley GK, Gorny KR, Henrichsen TL, Woodrum DA, Brown DL. A clinical review of focused ultrasound ablation with magnetic resonance guidance: an option for treating uterine fibroids. *Ultrasound Quarterly*. 2008; 24(no. 2):131–139. [PubMed: 18528271]
11. Cohen ZR, Zaubermaun J, Harnof S, Mardor Y, Nass D, Zadicario E, Hananel A, Castel D, Faibel M, Ram Z. Magnetic resonance imaging-guided focused ultrasound for thermal ablation in the brain: a feasibility study in a swine model. *Neurosurgery*. 2007; 60(no. 4):593–600. [PubMed: 17415195]
12. Kopelman D, Papa M. Magnetic Resonance-Guided Focused Ultrasound Surgery for the Noninvasive Curative Ablation of Tumors and Palliative Treatments: A Review. *Annals of Surgical Oncology*. 2007; 14(no.):1540–1550. [PubMed: 17318277]
13. Zhang S, Xia R, Lebrun L, Anderson D, Shrout TR. Piezoelectric materials for high power, high temperature applications. *Materials Letters*. 2005; 59(no. 27):3471–3475.
14. Chapelon JY, Cathignol D, Cain C, Ebbini E, Kluiwstra JU, Sapozhnikov OA, Fleury G, Berriet R, Chupin L, Guey JL. New piezoelectric transducers for therapeutic ultrasound. *Ultrasound in Medicine and Biology*. 2000; 26(no. 1):153–139. [PubMed: 10687803]
15. Zimmer JE, Hynynen K, He DS, Marcus F. The Feasibility of Using Ultrasound for Cardiac Ablation. *IEEE Transactions on Biomedical Engineering*. 1995; 42(no. 9):891–899. [PubMed: 7558063]
16. Rivens IH, Clark RL, ter Haar GR. Design of Focused Ultrasound Surgery Transducers. *IEEE Transactions on Ultrasonics, Ferroelectrics, and Frequency Control*. 1996; 43(no. 6):1023–1031.
17. Brentnall MD, Martin RW, Vaezy S, Kaczkowski P, Forster F F, Crum L. A new high intensity focused ultrasound applicator for surgical applications. *IEEE Transactions on Ultrasonics, Ferroelectrics, and Frequency Control*. 2001; 48(no. 1):53–63.
18. Ebbini E, Bischof JC, Coad JE. Lesion Formation and Visualization Using Dual-mode Ultrasound Phased Arrays. *Proceedings of the IEEE Ultrasonics Symposium*. 2001:1351–1354.
19. Gentry KL, Smith SW. Integrated catheter for 3-D intracardiac echocardiography and ultrasound ablation. *IEEE Transactions on Ultrasonics, Ferroelectrics, Frequency Control*. 2004; 51(no. 7):800–808.
20. Ergun AS, Huang Y, Zhuang X, Oralkan O, Yarahoglu GG, Khuri-Yakub BT. Capacitive micromachined ultrasonic transducers: fabrication technology. *IEEE Transactions on Ultrasonics, Ferroelectrics and Frequency Control*. 2005; 52(no. 12):2242–2258.
21. Oralkan O, Ergun AS, Johnson JA, Karaman M, Demirci Utkan, Kaviani K, Lee TH, Khuri-Yakub BT. Capacitive Micromachined Ultrasonics Transducer: Next-Generation Arrays for Acoustic Imaging? *IEEE Transactions on Ultrasonics, Ferroelectrics, and Frequency Control*. 2002; 49(no. 11):1596–1610.
22. Mills DM. Medical imaging with capacitive micromachined ultrasound transducer (cMUT) arrays. *Proc. IEEE Ultrasonics Symp*. 2004:384–390.
23. Caliano G, Carotenuto R, Cianci E, Foglietti V, Caronti A, Iula A, Pappalardo M. Design, fabrication and characterization of a capacitive micromachined ultrasonic probe for medical imaging. *IEEE Trans. Ultrason. Ferroelectr. Freq. Control*. 2005 Dec.52(no. 12):2259–2269. [PubMed: 16463491]
24. Degertekin FL, Guldiken RO, Karaman M. Annular-ring CMUT arrays for forward-looking IVUS: Transducer characterization and imaging. *IEEE Trans. Ultrason. Ferroelectr. Freq. Control*. 2006 Feb.53(no. 2):474–482. [PubMed: 16529123]
25. Kuo, Hung-Liang. Variation of ultrasonic velocity and absorption with Temperature and Frequency in High Viscosity Vegetable Oils”, *Jap. Journal of Applied Physics*. 1971; 10.(no. 2): 167–170.
26. Tables of Acoustic Properties of Materials. Available: <http://www.ondacorp.com>. (URL).

27. Huang Y, Haeggstrom EO, Zhuang X, Ergun AS, Khuri-Yakub BT. Optimized Membrane Configuration Improves CMUT Performance. *Proceedings of the IEEE Ultrasonics Symposium*. 2004:505–508.
28. Wong SH, Kupnik M, Zhuang X, Lin DS, Butts Pauly K, Khuri-Yakub BT. Evaluation of Wafer Bonded CMUTs with Rectangular Membranes Featuring High Fill Factor. *IEEE Transactions on Ultrasonics, Ferroelectrics and Frequency Control*. 2008; 55(no. 9):2053–2065.
29. Zhou S, Reynolds P, Hossack JA. Improving the Performance of Capacitive Micromachined Ultrasound Transducers using Modified Membrane and Support Structures. *Proceedings of the IEEE Ultrasonics Symposium*. 2005:1925–1928.
30. Huang Y, Haeggstrom EO, Zhuang X, Ergun AS, Khuri-Yakub BT. Capacitive micromachined ultrasonic transducers CMUTs with piston-shaped membranes. *Proceedings of the IEEE Ultrasonics Symposium*. 2005:594–598.
31. Senlik MN, Olcum S, Atalar A. Improved Performance of CMUT with Nonuniform membranes. *Proceedings of the IEEE Symposium on Ultrasonics*. 2005:597–600.
32. French PJ, Sarro PM. Surface versus bulk micromachining: the contest for suitable applications. *Journal of Micromechanical Engineering*. 1998; 8(no. 2):45–53.
33. Huang Y, Haeggstrom EO, Zhuang X, Ergun AS, Khuri-Yakub BT. A Solution to the Charging Problem in Capacitive Micromachined Ultrasonic Transducers. *IEEE Transactions on Ultrasonics, Ferroelectrics, and Frequency Control*. 2005; 52(no. 4):578–580.
34. Kupnik M, Ergun AS, Huang Y, Khuri-Yakub BT. Extended insulation layer structure for CMUTs. *Proceedings of the IEEE Ultrasonic Symposium*. 2007:511–514.
35. Ladabaum I, Jin XC, Soh HT, Atalar A, Khuri-Yakub BT. Surface micromachined capacitive ultrasonic transducers. *IEEE Transactions on Ultrasonics, Ferroelectrics and Frequency Control*. 1998; 45(no. 3):678–690.
36. Ebbini ES, Yao H, Shresha A. Dual-Mode Ultrasound Phased Array for Image-guided Surgery. *Ultrasonic Imaging*. 2006; 28(no. 2):201–220.
37. Wong, SH.; Ergun, AS.; Yaralioglu, GG.; Oralkan, O.; Kupnik, M.; Butts, RK.; Khuri-Yakub, BT. Design of HIFU CMUT Arrays for Treatment of Liver and Renal Cancer; AIP Conference Proceedings, Therapeutic Ultrasound: 6th International Symposium on Therapeutic Ultrasound; 2007. p. 54-60.
38. Wong SH, Kupnik M, Butts-Pauly K, Khuri-Yakub BT. Advantages of capacitive micromachined ultrasonics transducers (CMUTs) for high intensity focused ultrasound (HIFU). *Proceedings of the IEEE Ultrasonics Symposium*. 2007:1313–1316.
39. Ergun AS, Barnes S, Gardner E. An Assessment of the Thermal Efficiency of Capacitive Micromachined Ultrasonic Transducers. *Proceedings of IEEE Ultrasonics Symposium*. 2007:420–424.
40. Chen Y, Wen Y, Lai P. Loss Mechanisms in Piezoelectric Transducer and Its Response to Stress. *IEEE Information Acquisition*. 2004:213–219.
41. Freer, J.; Kutty, K.; Nam, H.; Wee, K.; Woodard, R. *Materials Handbook for Hybrid Microelectronics*. Boston, MA: Artech House; 1988.
42. Martin RW, Vaezy S, Proctor A, Myntti T, Lee JB, Crum LA. Water-cooled High-Intensity Ultrasound Surgical Applicators with Frequency Tracking. *IEEE Transactions on Ultrasonics, Ferroelectrics, and Frequency Controls*. 2003; 50(no. 10):1305–1326.
43. Lafon C, Chapelon JY, Prat F, Gorry F, Margonari J, Theillre Y, Cathignol D. Design and preliminary results of an ultrasound applicator for interstitial thermal coagulation. *Ultrasound in Medicine and Biology*. 1998; 24(no. 1):113–122. [PubMed: 9483778]
44. Berlincourt, D.; Krueger, HHA.; Near, C., editors. Technical Publication TP-226 Properties of Piezoelectricity Ceramics. Available: <http://morganelectroceramics.com>. (URL)
45. Schenck JF. The role of magnetic susceptibility in magnetic resonance imaging: MRI magnetic compatibility of the first and second kinds. *Medical Physics*. 1996; 23(no. 6):815–850. [PubMed: 8798169]
46. Wong SH, Watkins RD, Kupnik M, Butts-Pauly K, Khuri-Yakub BT. MRI Temperature Monitoring of HIFU CMUTs. *IEEE Transactions on Ultrasonics, Ferroelectrics, and Frequency Control*. 2008; 55(no. 4):811–818.

47. Khuri-Yakub BT, Cheng CH, Degertekin FL, Ergun AS, Hansen ST, Jin XC, Oralkan O. Silicon Micromachined Ultrasonic Transducers. *Japanese Journal Applied Physics*. 2000; 39(no. 2):2883–2887.
48. Eccardt K PC, Niederer. Micromachined ultrasound transducers with improved coupling factors from a CMOS compatible process. *Ultrasonics*. 2000; 38:774–780. [PubMed: 10829770]
49. Zhuang X, Ergun AS, Huang Y, Wygant IO, Oralkan O, Khuri-Yakub BT. Integration of trench-isolated through-wafer interconnects with 2D capacitive micromachined ultrasonic transducer arrays. *Sensors and Actuators A: Physical*. 2007; 138(no. 1):221–229.
50. Wong SH, Kupnik M, Watkins RD, Butts Pauly K, Khuri-Yakub BT. Progress in CMUTs for HIFU ablation of upper abdominal cancer. *Proceedings of IEEE Ultrasonics Symposium*. 2008:2068–2071.
51. Habash, R. Bioeffects and therapeutic Applications of Electromagnetic Energy. Boca Raton, FL: CRC Press; 2008.
52. O'Neil HT. Theory of focusing radiators. *Journal of the Acoustical Society of America*. 1949; 21(no. 5):516–526.
53. Hutchinson EB, Buchanan MT, Huynynen K. Design and optimization of an aperiodic ultrasound phased array for intracavitary prostate thermal therapies. *Medical Physics*. 1996; 23(no. 5):767–776. [PubMed: 8724752]
54. Macovski, A. Medical Imaging Systems. Upper Saddle River, NJ: Prentice Hall; 1983.
55. Goss SA, Frizzell LA, Dunn F. Ultrasonic Absorption and Attenuation in Mammalian Tissues. *Ultrasound in Medicine and Biology*. 1979; 5(no. 2):181–186. [PubMed: 556199]
56. Pennes HH. Analysis of tissue and arterial blood temperature in the resting forearm. *Journal of Applied Physiology*. 1948; 1:93–122. [PubMed: 18887578]
57. Sapareto SA, Dewey WC. Thermal dose determination in cancer therapy. *International Journal of Radiology Oncology Biology Physics*. 1984; 10:787–793.
58. Yaralioglu GG, Ergun AS, Khuri-Yakub BT. Finite-element analysis of capacitive micromachined ultrasonics transducers. *IEEE Transactions on Ultrasonics, Ferroelectrics, and Frequency Control*. 2005; 52(no. 12):2185–2198.
59. The ANSYS Element Reference". Available: <http://www.gidb.itu.edu.tr/staff/unsan/B/AnsysBook/Element.pdf>. (URL).
60. Fahy, FJ. Foundations of engineering acoustics. 2nd ed.. London, UK: Academic Press; 2003.
61. Huang Y, Haeggstrom E, Bayram B, Zhuang X, Ergun AS, Cheng CH, Khuri-Yakub BT. Fabricating capacitive micromachined ultrasonics transducers with wafer-bonding technology. *Microelectromechanical Systems*. 2003; 12(no. 2):128–137.
62. Kino, G. Acoustic waves: devices, imaging, and analog signal processing. Englewood Cliffs, N.J.: Prentice-Hall; 1987.
63. HNP Hydrophone. Available: <http://www.ondacorp.com>. (URL).
64. Bayram B, Kupnik M, Yaralioglu GG, Oralkan O, Ergun AS, Lin D, Wong SH, Khuri-Yakub BT. Finite element modeling and experimental characterization of crosstalk in 1-D CMUT arrays. *IEEE Transactions on Ultrasonics, Ferroelectrics and Frequency Control*. 2007; 54(no. 2):418–430.
65. Wilm M, Reinhardt A, Laude V, Armati R, Daniau W, Ballandras S. Three-dimensional modelling of micromachined-ultrasonic-transducer arrays operating in water. *Ultrasonics*. 2005; 43(no. 6): 457–465. [PubMed: 15823320]
66. Glorieux, Christ; Van de Rostyne, K. On the character of acoustic waves at the interface between soft solids and liquids. *Journal of the Acoustical Society of America*. 2001; 110(no. 3):1299–1306. [PubMed: 11572340]
67. Zhu J, Popovics J. Leaky Rayleigh and Scholte waves at the fluid-solid interface subjected to transient point loading. *Journal of the Acoustical Society of America*. 2004; 116(no. 4):2101–2110.
68. Dias JF. An experimental investigation of the cross coupling between elements of an acoustic imaging array transducer. *Ultrasonic Imaging*. 1982; 4:1137–1139.

69. Yaralioglu GG, Bayram B, Khuri-Yakub BT. Finite Element Analysis of CMUTs: Conventional vs. Collapse Operation Modes. *Proceedings of the IEEE Ultrasonics Symposium*. 2006:586–589.
70. Stengl R, Ahn KY, Gosele U. Bubble-free Silicon Wafer Bonding in a Non-Cleanroom Environment. *Japanese Journal of Applied Physics*. 1990; 27:L2364–L2366.
71. Esser RH, Hobart KD, Kub FJ. Improved Low-Temperature Si-Si Hydrophilic Wafer bonding. *Journal of the Electromechanical Society*. 2003; 150(no. 3):228–231.
72. Roh Y, Khuri-Yakub BT. Finite element analysis of underwater capacitor micromachined ultrasonic transducers. *IEEE Transactions on Ultrasonics, Ferroelectrics, and Frequency Control*. 2002; 49(no. 3):293–298.
73. Bayram B, Kupnik M, Yaralioglu GG, Oralkan O, Ergun AS, Lin DS, Wong SH, Khuri-Yakub BT. Finite Element Modeling and Experimental Characterization of Crosstalk in 1-D CMUT Arrays. *IEEE Transactions on Ultrasonics, Ferroelectrics, and Frequency Control*. 2007; 56(no. 2): 418–430.

Biographies

Serena Wong was born in Stanford, CA. She received her B.S. and M.S. degrees in electrical engineering from Stanford University, Stanford, CA, in 2002 and 2003, respectively. She is currently a Ph.D. candidate in electrical engineering at the E. L. Ginzton Laboratory of Stanford University. She is a 2002 National Science Foundation Fellow, 2002 Stanford Graduate Fellow, and a member of Tau Beta Pi.

From 1998 through 2001, she worked as a research assistant in the Pre-polarized Magnetic Resonance Imaging (pMRI) laboratory at Stanford University in an effort to develop a cost-effective MRI system. This system would greatly reduce costs of imaging technology and improve accessibility for patients. Recently, she has been interested in high intensity focused ultrasound (HIFU) as a minimally-invasive therapeutic tool for treatment of cancer. She is presently working with capacitive micromachined ultrasonic transducers (CMUTs) for HIFU applications such as the treatment of lower abdominal cancers.

Mario Kupnik is currently a research associate in electrical engineering at Edward L. Ginzton Laboratory at Stanford University. He received his Ph.D. from the University of Leoben, Austria, in 2004, and the Diplom Ingenieur degree from Graz University of Technology, in 2000. From summer 1999 to October 2000, he worked as an Analog Design Engineer for Infineon Technologies AG, Graz, on the design of ferroelectric memories and contactless smart card (RFID) systems. His present research interests include the design, modeling, fabrication, and application of micromachined sensors and actuators, with a main focus on capacitive micromachined ultrasonic transducers mainly for air-coupled applications, including high gas temperatures. Examples are transit-time gas flowmeters for hot and pulsating gases, ultrasonic nondestructive evaluation using non-contact ultrasound, nonlinear acoustics, and bio/chemical gas sensing applications (electronic nose). Dr. Kupnik

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Ronald D. Watkins is a Senior Scientific Research Associate in the Stanford School of Medicine and works in the Richard M. Lucas Center for Radiological Sciences and the Edward L. Ginzton Lab of Applied Physics. Before joining Stanford, he worked at General Electric Medical Systems in Rancho Cordova Ca. and GE Corporate Research in Niskayuna N.Y. Early in his career he focused on digital X-Ray image processing, CT Data Acquisition, and MRI subsystems. He then moved to Ultrasound imaging systems and in 1991 and on he developed systems for MR Guided Focused Ultrasound Surgery and Very High Field MR Imaging Systems. He has 37 issued US Patents and 46 external publications and conference proceedings. He is a member of the IEEE and the ISMRM.

Kim Butts-Pauly Kim Butts Pauly received a B.S. degree in physics from Duke University, Durham, NC. She received her Ph.D. in biophysical sciences from the Mayo Graduate School in Rochester, MN. In 1996, she joined the faculty at Stanford University's Department of Radiology, where she has been working on MR-guided high intensity focused ultrasound, cryoablation, RF ablation, and biopsy. She has also been involved with the development of a truly integrated X-Ray and MRI system.

Butrus T. (Pierre) Khuri-Yakub is a Professor of Electrical Engineering at Stanford University. He received the BS degree in 1970 from the American University of Beirut, the MS degree in 1972 from Dartmouth College, and the Ph.D. degree in 1975 from Stanford University, all in electrical engineering. He was a Research Associate (1965–1978) then Senior Research Associate (1978–1982) at the E. L. Ginzton Laboratory of Stanford University and was promoted to the rank of Professor of Electrical Engineering in 1982. His current research interests include medical ultrasound imaging and therapy, micromachined ultrasonic transducers, smart bio-fluidic channels, microphones, ultrasonic fluid ejectors, and ultrasonic nondestructive evaluation, imaging and microscopy. He has authored over 400 publications and has been principal inventor or co-inventor of 76 US and International issued patents. He was awarded the Medal of the City of Bordeaux in 1983 for his contributions to Nondestructive Evaluation, the Distinguished Advisor Award of the School of Engineering at Stanford University in 1987, the Distinguished Lecturer Award of the IEEE UFFC society in 1999, a Stanford University Outstanding Inventor Award in 2004,

and a Distinguished Alumnus Award of the School of Engineering of the American University of Beirut in 2005.

Fig. 1. Photograph of the 8-element concentric ring array. The array has front side traces and bonding pads.

< constrained in x ^ constrained in y

Fig. 2.

Schematic of the wave-guide, axisymmetric CMUT cell model simulated in ANSYS. The CMUT silicon and oxide are modeled using PLANE42 elements, while the water column is modeled with FLUID29 elements. TRANS126 elements in the gap of the cell convert electrical to mechanical energy. The boundary conditions are set so the bottom and side of the wafer are immovable in y and x, respectively. The average output pressure in harmonic and dynamic analysis is calculated half a wavelength from the surface of the cell and averaged.

Fig. 3.

Simplest process for a wafer-bonded CMUT. A first oxidation (a) and oxide etch defines the cavity (b). A second oxidation forms the insulation layer; the second oxide etch removes the bulged edges from oxidizing a nonplanar surface (see circles in c). A fusion bonding step is used to bond the first wafer to a SOI wafer and the handle wafer is removed (d). Finally the ground and signal connections are defined (e) and the array elements are separated (f).

Fig. 4.

Setup used to measure the pressure of the CMUT in immersion with a needle hydrophone. These measurements were used to calculate the surface acoustic output pressure of the device after compensating for the frequency response of the hydrophone and acoustic attenuation and diffraction to determine the frequency and acoustic response of the device.

Fig. 5.

Block diagram of electronics for the ring array. Direct digital synthesizers control the frequency and phase of each channel. These signals are then amplified and superimposed with a DC signal using a bias T. The signal is then applied to each transducer element, which is matched to 50 ohms with an series inductor. While the system is easily expandable to 8 channels, since only four elements were active, we only show the electronics connections for four channels in this figure.

Fig. 6. Simulated beam profile in homogenized liver of the 8-element concentric ring array in the focal plane (30dB dynamic range) (a) and along the axial direction (b) 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/cm^2) will produce a focal intensity of 680 W/cm^2 .

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 seconds.

Fig. 8.

Schematic of the circular cells used in our design. The cells were patterned in a close packed arrangement (a) into both test transducers and concentric ring arrays. The cross section of an individual cell (b) 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.

Fig. 10.

Comparison of the measured and simulated dynamic response of the HIFU design with DC voltage of 172 V (70% of pull-in voltage) and various AC voltages. The deviation between measured and simulated occurs because of charging in the oxide layer and acoustic crosstalk.

Fig. 11.

Simulated beam profile of the center 4 elements in the 8-element concentric ring array focused in oil (a). The measured profile with 30dB dynamic range (b) shows less symmetry than the simulated profile because of the front side traces. The relationship of the measurement plane to the surface of the transducer is shown in (c). Beam profile (d) through the center of the concentric ring array shows that the measured beamwidth is 10% narrower than the calculated beamwidth. The side lobes of the measured profile are also two times

greater in magnitude than the simulated profile. These deviations are caused by acoustic crosstalk.

TABLE I

Material data used for simulation

-	soybean oil	homogenized liver	blood
speed of sound (m/s)	1430	1550	-
density (kg/m ³)	930	1060	1000
attenuation (dB/cm)	$0.47f^{1.12}$	$0.08f^{1.85}$	-
heat capacity (J/kg-K)	-	3600	3840
conductivity (W/K-m) -	0.5	-	
blood perfusion rate (kg/m ³ /s)	-	-	0.5