CAPACITIVE MICROMACHINED ULTRASONIC TRANSDUCER (CMUT) FOR MR-GUIDED NONINVASIVE THERAPEUTIC ULTRASOUND APPLICATIONS

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ABSTRACT

In the last decade, noninvasive therapeutic ultrasound guided by magnetic resonance imaging (MRI) has increased in popularity for diseases such as cancer and heart arrhythmias. While piezoelectric transducers are the dominant technology for therapeutic ultrasound applications, recently capacitive micromachined ultrasonic transducers (CMUTs) have demonstrated competitive performance and ease of fabrication. In this paper, we present a CMUT design for MR-guided therapeutic ultrasound, with circular cells and center mass (piston) that fulfills the requirements for therapy of upper abdominal cancers, i.e. 1 MPa peak to peak output pressure at 2.5 MHz. In particular, we discuss the choice of cell shape and membrane topography and their influences on output pressure of the device.

KEYWORDS

HIFU, CMUT, MR-guidance, therapeutic ultrasound.

INTRODUCTION

Noninvasive therapeutic ultrasound, guided by magnetic resonance imaging (MRI), potentially reduces mortality, lowers costs, and increases accessibility of treatments when compared with traditional surgeries. Typically, piezoelectric transducers are used for therapeutic ultrasound, but recently capacitive micromachined ultrasonic transducers (CMUTs) have become competitive, providing advantages in performance, reliability, and fabrication. MRI is critical to treatment planning and real-time thermal feedback for therapeutic ultrasound treatments (Fig. 1).

Requirements for MR compatibility and reduction of image artifacts require transducer designs that avoid large metal electrodes. A second requirement is continuous wave (CW) operation at a high acoustic output pressure because treatments are applied from outside the body for long periods of time. For treatment of upper abdominal cancers, 1 MPa peak to peak output pressure at 2.5 MHz is necessary to treat 2-3 cm sized tumors within 1 hour [1].

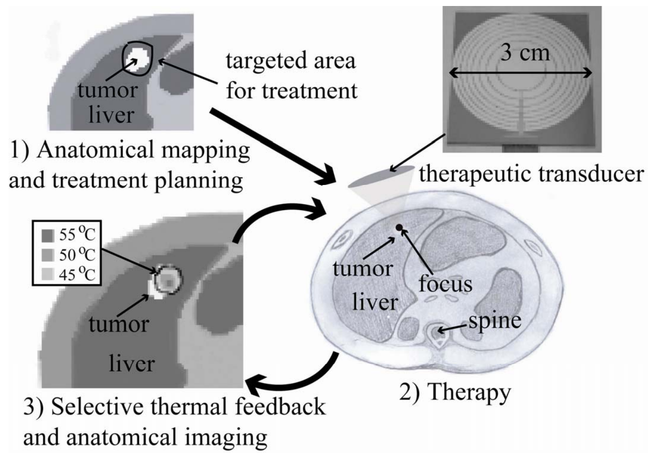

Figure 1: MRI is used to plan the targeted area for treatment around the tumor (1). A CMUT 8-element annular ring array (2) focuses ultrasound to heat and kill cancerous cells; heating constantly monitored with thermal feedback from MRI to adjust the therapy in real-time (3).

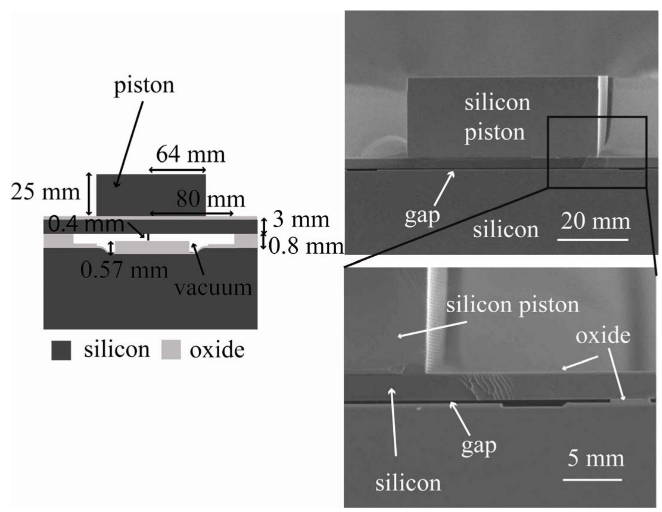

Figure 2: Schematics (a) and SEMs (b) of a circular single cell of a piston CMUT. . The transducer is able to produce CW 1 MPa peak to peak output pressure used for the therapeutic ultrasound applications in an 8-element concentric ring array.

In this paper, we present a CMUT- based design with 80 micron radius, circular-shaped, conductive silicon membranes that feature 64 micron radius silicon pistons on the membrane surface (Fig.

2). Compared to existing work [2, 3], this approach provides advantages in process simplicity, output pressure performance, and compatibility with MR-imaging systems. The circular shape is less sensitive in terms of excitation of higher order modes due to acoustic crosstalk waves at the transducer surface and the increased stiffness of the silicon piston features higher average membrane displacements. Both advantages allow achieving high acoustic output pressures in CW. In addition, because conductive silicon is used for the membrane material, no metal electrodes are required, which makes the devices compatible with MRI.

METHODS

Design and Fabrication

We use a direct wafer bonding-based process for the realization of the CMUT, as shown in Figure 2. Compared to the more traditional sacrificial-release process [4], this enables the fabrication of large cells with variable membrane topography, such as the piston shaped membranes. Large sized cells and variable membrane topography are needed for the lower frequencies and output pressures needed for high intensity focussed ultrasound (HIFU) applications. To fill the requirements for MR compatibility, CMUT cells with entirely conductive membranes were fabricated to reduce the large sheets of metal on the surface of the device that would cause MR artifacts [5].

We focused on the two main design parameters to achieve the requirements of high output pressure, the cell shape and the piston on top of the silicon membrane. We compared the output pressure of a 70-micron radius circular cell to a 110 micron by 550-micron rectangular cell. Both cells had a membrane thickness of 6 micron, a gap height of 0.4 microns, and an insulation thickness of 0.6 microns. These cells had similar pull-in voltages (140 V) and center frequencies (3 MHz). A center piston was used to maximize both average center displacement of the membrane and output pressure. We compared the circular-shaped piston device, with dimensions shown in Figure 2, to a non-piston cell device with similar center frequency and pull-in voltage, a circle with 80 micron radius, a membrane thickness of 6 micron, a gap height of 0.4 microns, and an insulation thickness of 0.6 microns.

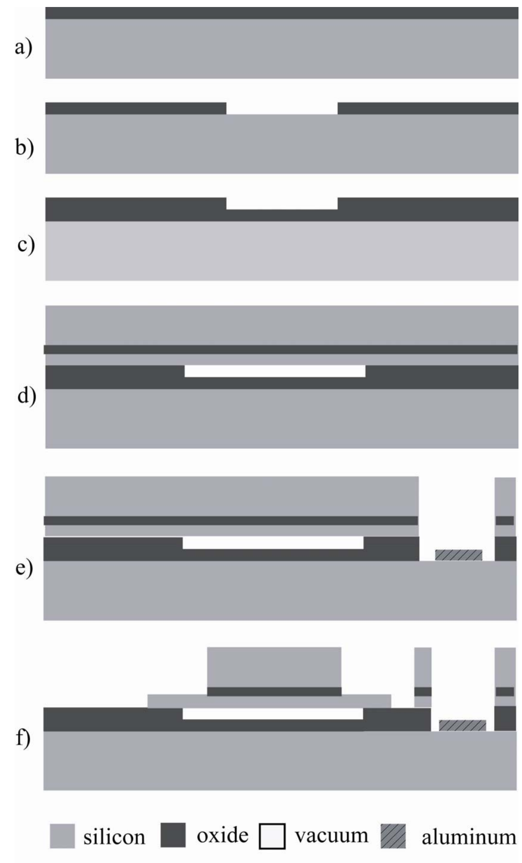

Figure 3: The prime wafer is first oxidized (a) and then etched (b) to form the cavities. A second oxidation (c) forms the insulation layer that protects the device from dielectric breakdown. In the case of a piston device, a double SOI wafer is fusion bonded to the cavity wafer (d) and the handle and buried oxide layer are removed by wet etching.

In the wafer bonding process [6], first cavities are realized on one side of a heavily doped and oxidized silicon prime wafer. First, the oxide layer (Fig. 3(a)) is patterned. The silicon beneath the oxide acts as an etch stop for this step (Fig. 3(b)). Then, the wafer is oxidized again to form the bottom insulation layer in the cavity of the each cell (Fig. 3(c)). This allows excellent vertical control of the device dimensions. For a simple, non-piston cell, a silicon-on-insulator wafer (SOI) is then bonded to the substrate wafer using direct fusion bonding (Fig. 3(d)), and then the handle wafer and buried oxide layer (BOX) are removed. For the piston shape design a double SOI wafer with the desired piston and membrane thicknesses is used. Then the ground connections and electrodes are etched into the wafer and metal pads are deposited (Fig. 3(e)). In the final step, pistons shapes are etched in the top layer of silicon in the double SOI wafer (Fig.3(f)])

Measurements Methods

We measured the dynamic output pressure and membrane displacement to evaluate the performance of different devices. The 2.5 mm by 2.5 mm square test transducers were mounted on a printed circuit board using conductive epoxy. Gold wire bonds were used to connect the pads on the device with the printed circuit board traces.

Dynamic displacement in immersion was measured using an optical laser interferometer (Polytec OFV511, Polytec Corporation, Tustin, CA) (Fig. 4(a)). The devices were immersed in soybean oil for electrical isolation. The acoustical properties of soybean oil are very similar to human tissue [7]. A 30-cycle, tone-burst excitation at 2.5 MHz the center frequency of the CMUT in oil, was superimposed with a DC voltage 80% of the pull-in voltage. We used the laser interferometer to measure the displacement over time for spatial locations every 20 microns over a 1.5-by-1.5 mm area, which corresponds to one-quarter of CMUT. From these displacement-over-time signals, we generated time-dependent 3-D surface measurements of the displacement. We compensated for the index of refraction of oil, 1.47 [8], to calculate the displacement of the front transducer surface.

For dynamic output pressure measurements, the CMUTs were immersed in a tank filled with soybean oil (Fig. 4 b) and a PZT Z44 0400 hydrophone (Onda Corporation, Sunnyvale, CA) was positioned 2 cm from the surface. The measurement data was corrected for the frequency response [9] of the hydrophone and sound attenuation and diffraction [10], which allows us to calculate the acoustic output pressure at the surface of the transducer.

RESULTS

Cell shape

Although high fill factor cells, such as rectangular-shaped cells, seem appealing for increasing average output pressure [11], we found that in CW operation at a single frequency, dispersive guided modes influence the mode shape. Instead of deflecting synchronously, the rectangular cells deflect asynchronously in the lengthwise direction (Fig. 5(a)). The result is a reduced average output displacement and output pressure. The rectangular cells deflect asynchronously in the lengthwise direction (Fig. 5(a)), which results in a reduced

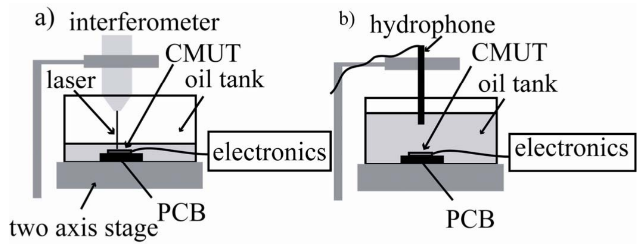

Figure 4: Setups for measuring dynamic displacement (a) and acoustic output pressure (b).

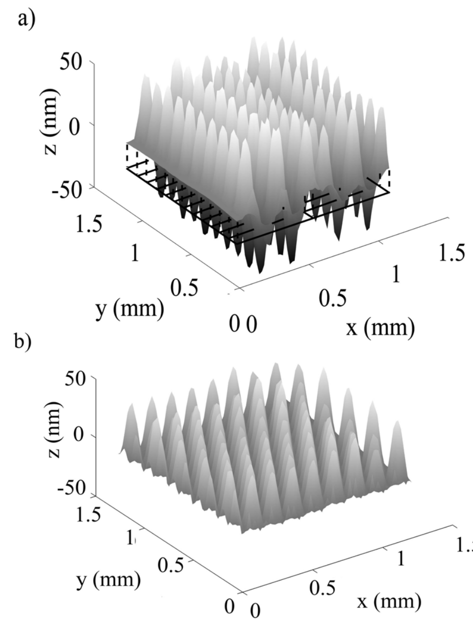

Figure 5: Displacement during steady state operation for a rectangular (a) versus a circular (b) CMUT design.

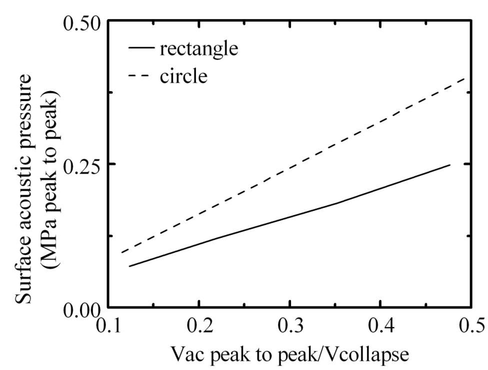

Figure 6: Acoustic output pressure of a circular cell design compared to a rectangular cell design e with comparable pull-in voltages and center frequencies.

average output displacement and output pressure. The circular cells, however, demonstrate synchronous displacement (Fig. 5(b)). The lowest frequency of higher order modes of the circular cells is 2.2 times higher than the fundamental frequency, while for a rectangular cell of aspect ratio 1:5, this ratio is only 1.18 [12]. When comparing a circle and rectangular cell (aspect ratio 1:5) with similar pull-in voltages and center frequencies, the circular-shaped cells outperform the rectangular-shaped cells in CW mode (Fig. 6), which is an essential result to know for the application of CMUTs in HIFU applications.

Membrane Configuration

Adding a silicon piston to the center of each membrane increases the average output displacement and output pressure (Fig. 2). Compared to other piston designs [13,14], our design eliminates the need for metal as the top electrode, which makes it ideal for use under MRI. In addition, compared to [13], our process provides full insulation coverage of the bottom electrode, which is essential to avoiding electrical breakdown from high voltages. Third, the layer of oxide separating the silicon membrane and silicon piston layers, formed by direct wafer bonding of two silicon-on-oxide (SOI) wafers, acts as an etch stop that improves fabrication uniformity across the large-area (3 cm in diameter) transducer array. Finally, compared to [13], our process reduces the number of lithography steps. Compared to non-piston designs with similar center frequency, our piston transducer delivers 1.3 times higher acoustic output pressure (Fig. 7). In addition, we demonstrate successful CW operation at 1 MPa peak to peak for 1 hour, which allows us to use this transducer for therapeutic ultrasound applications.

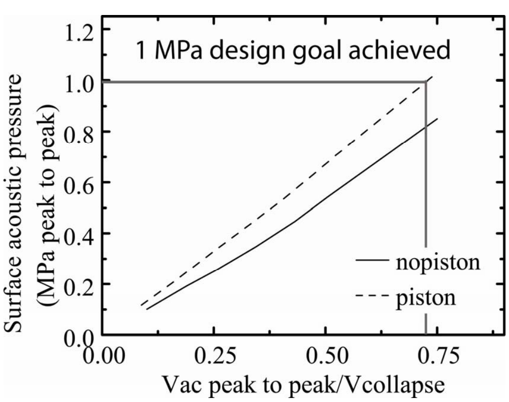

Figure 7: Output pressure of a piston CMUT compared to a non-piston device.

CONCLUSIONS

Using a circular, silicon piston CMUT cell, we fulfill the needs of high output pressure and MR compatibility for noninvasive image-guided therapy. CMUTs that feature circular shaped cells with silicon pistons fulfill the challenging requirements of high acoustic output pressure in CW mode operation for noninvasive image-guided therapy application. The circular cell shape and piston-shape increases the transducer's average output pressure. Furthermore, the design allows us to eliminate metal on the transducer surface, which significantly reduces MR artifacts and, thus, opens the door to excellent MR compatibility. Our transducer design demonstrates significant advantages over current existing technologies because of the simplification of the fabrication process and the elimination of metals from the transducer's surface. With this transducer design, we will develop full concentric ring arrays for noninvasive therapeutic treatment of upper abdominal cancers.

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