

CAPACITIVE MICROMACHINED ULTRASONIC TRANSDUCERS FOR HIGH INTENSITY FOCUSED ABLATION OF UPPER ABDOMINAL TUMORS

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Abstract – We present the development of a capacitive micromachined ultrasonic transducer (CMUT) array for noninvasive focused ultrasound ablation of lower abdominal cancers under MR-guidance. While piezoelectric transducers have been traditionally used for high intensity focused ultrasound (HIFU), recent advances in capacitive micromachined ultrasonic transducers (CMUTs) have made them highly competitive with regard to costs, fabrication flexibility, and performance. Even current imaging CMUTs have shown capability of HIFU operation through high power and continuous wave operation. In this paper, we will show our experiments with current imaging CMUTs operated in HIFU mode. In addition, we will show the design and development of CMUT membranes and a transducer array specifically for HIFU ablation lower abdominal cancers.

I. INTRODUCTION AND BACKGROUND

Neoplasms of the liver, such as metastatic colorectal cancer and hepatocellular carcinoma, cause significant morbidity and mortality [1]. Resection of these cancers improves five-year survival from 8% to 30-35% [2]–[4]. Unfortunately, only 20% of patients are suitable for resection [5]. For 80% of patients, a potentially curative non-invasive treatment option, such as MR-guided high intensity focused ultrasound (HIFU) therapy, is needed. While tabletop MR-guided, HIFU systems have been developed, these systems have the transducer fixed in the MR scanner bed, which limits the acoustic windows available [7].

Traditionally, piezoelectric transducers have been used for HIFU [8]. The performance of piezoelectric transducers is limited by their material properties. Piezoelectric transducers are inherently high impedance (30 MRayl) compared to tissue (1.5 MRayl). This means that often complex backing and matching layers have to be designed to achieve reasonable bandwidth and efficient energy transfer to the medium [9], [10]. Heating from the dielectric loss tangent is often worsened by these insulating layers and can often cause the transducer to depole. Fabrication of piezoelectric devices of certain shapes and sizes and integration with electronics can also be a challenge.

The CMUT membrane structure allows a large bandwidth and efficiency because the membrane impedance is small compared to that of the medium. CMUTs also allow cost-effective and flexible processing; arrays from 100 μm to 5.6 cm sizes with frequencies between 10 kHz - 60 MHz have been demonstrated [11], [12], [14]–[18]. Silicon is also more thermally conductive and may not suffer from as large

thermal heating issues as piezoelectric transducers.

In this paper, we will show the promising performance capabilities of typical CMUT imaging transducers. With these promising results, we have designed and simulated a HIFU CMUT membrane and hand-held, annular ring probe for use under MR guidance. This handheld probe will be mechanically moveable and not limited to the acoustic windows of a fixed transducer in a MR scanner bed.

II. EXPERIMENTS WITH CURRENT IMAGING TRANSDUCERS

In order to demonstrate the potential of CMUTs for HIFU, we used typical imaging transducers in high power and continuous wave mode. These CMUTs have demonstrated output surface pressures as high as 1 MPa peak-to-peak and survived continuous wave (CW) operation for times beyond 25 minutes.

A. Methods and Setup

A 1.8 by 0.66 mm imaging CMUT was placed in vegetable oil, which mimics the acoustic behaviour of human tissue and also provides electrical isolation in contrast to water. A hydrophone (Z44_0400, Onda Corporation, Sunnyvale, CA) was placed 1 cm in front of the transducer to measure the acoustic output pressure, which was corrected for attenuation and diffraction [19] to obtain the average pressure at the surface of the CMUT (Fig. 1).

We applied various signals to the CMUT to test high power and continuous wave (CW) operation. To test voltage versus pressure output, the transducer was excited by 3 MHz, 10 cycle, sinusoidal tone burst. The applied DC voltages were swept from 100-180 V, and AC voltages were selected so that the overall voltage did not exceed 250 V to prevent dielectric breakdown of the silicon dioxide isolation later. Continuous wave sinusoidal signals lasting for 1 min to several hours were also applied to the CMUT.

B. Results

The DC and AC voltage were swept and plotted against surface pressure (Fig. 1). Though these CMUTs were designed for imaging at 7 MHz, they could generate output pressures as high as 1 MPa peak to peak at 3 MHz, even when operated greatly off resonance. When the CMUT is biased near the collapse voltage and excited with a large

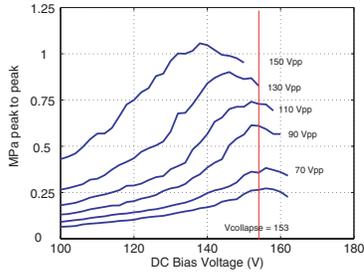


Fig. 1. Measurement result of the acoustic pressure, obtained by a hydrophone and corrected for diffraction and attenuation as a function of DC and AC bias voltage.

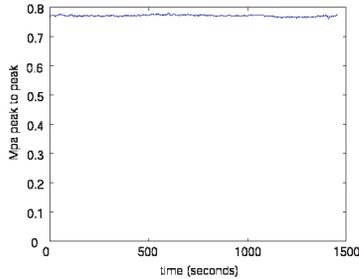


Fig. 2. Hydrophone measurements of continuous wave operation of a typical imaging CMUT.

enough AC voltage, the membrane can displace nearly the entire extent of the gap, which maximizes output pressure.

The same device was then operated in CW operation for 25 min with a continuous output pressure of 0.77 MPa peak-to-peak (Fig. ??). The major limitations of high power CW operation are the breakdown of the insulating layers from the high voltages and the failure of metal traces from high current densities. These issues can be solved by increasing the thickness of the insulation layer and interconnects, with further room for performance improvement.

III. DESIGN AND SIMULATION OF HIFU ANNULAR RING

With the promising results shown by the imaging CMUTs, we have designed and simulated new CMUT membranes specifically for HIFU. These membranes are characterized by their piston-like shape, which increases average displacement and output pressure. Using these newly designed membranes, we designed an annular array for non-invasive treatment of liver and renal cancers. This array could produce a 5 mm region of necrosis in 20 seconds in a homogeneous region of liver tissue.

A. CMUT Membrane Design and Simulation

Since non-invasive HIFU needs to penetrate through thick tissue, we designed our CMUT membranes to have lower center frequencies (1-3 MHz). Because HIFU depends on the power gain of the array design, a surface output pressure of 1.5-2.5 MPa peak to peak output pressure

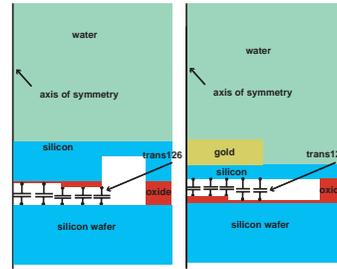


Fig. 3. ANSYS FEM models used to simulate the piston CMUT membranes formed by a double fusion bonding process (left) and electroplating process (right).

will be adequate to achieve the intensities necessary for ablation.

These new membrane designs are characterized by their piston-like shape. The piston is created by fabricating a thick mass in the center of the membrane, either by electroplating a heavy gold electrode or by creating a thick layer of silicon using two fusion bonding steps [12]. The main purpose of using a piston-shaped membrane is to be able to control the mass and spring constant independently. Additionally, a piston-shaped membrane provides an increased average membrane displacement, which increases the output pressure. The drawback is that the extra mass and stiffness reduces bandwidth. This does not negatively affect the transducer's performance when used for HIFU applications, because of their single frequency excitation.

Our new membrane design is also characterized by thicker electrode traces to minimize the current density and destruction of traces from high power. Also, a post design was used, where a stair of oxide is cut out from the main insulation layer. This minimizes the oxide charging effects and decreases the electric field in the oxide, which prevents dielectric breakdown and increases reliability [13].

1) *Simulation Methods:* Designs were modeled in ANSYS (ANSYS, Inc, Canonsburg, PA) assuming a constant temperature of 37 °C throughout the structure and no residual stress in the materials. The CMUT oxide, silicon, and metals were modeled using PLANE42 elements (Fig. 3). Circular membranes were simulated with axisymmetric symmetry, while rectangular membranes were simulated with planar symmetry and assumed to be infinite in length. Trans126 elements were used to simulate the transfer from voltage to mechanical force. A non-attenuating, non-absorbing water column of FLUID29 elements three wavelengths high was constructed on top of the CMUT, and an absorbing boundary was placed three wavelengths away from the surface. The pressure was measured half a wavelength away from the surface and averaged. Since the FLUID29 elements were non-absorbing, the average pressure calculated is the equivalent to the average surface pressure.

2) *Simulation Results:* Designs of several widths, membrane thicknesses, gap heights, and different geometries

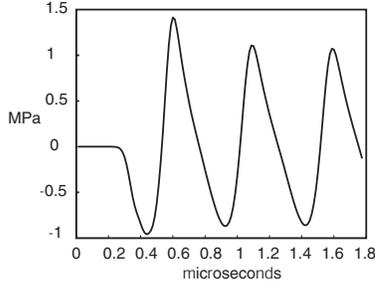


Fig. 4. Transient simulation result of the output pressure from a rectangular silicon piston CMUT design at 2 MHz. The collapse voltage of the transducer is 150 V. A 123 V DC voltage was applied and a 130 VPP AC voltage was applied on top of that.

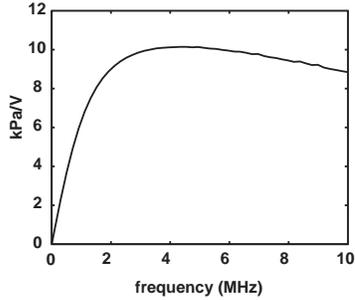


Fig. 5. Harmonic response for rectangular piston CMUT biased to 80% of the collapse voltage (150V).

have been selected for fabrication. These designs have center frequencies ranging from 1-4 MHz and output pressures ranging from 0.75 MPa peak-to-peak to greater than 2 MPa peak-to-peak. Example dynamic and frequency responses are shown in Fig. 4 and 5.

The best designs have piston widths between 60-85% of the membrane width and a piston thickness of at least 5 times the thickness of the membrane. This piston width allows a flatter, larger, more piston-like membrane while still keeping a reasonable collapse voltage and good separation of elasticity and mass.

Gold piston designs showed slightly higher output pressures and narrower bandwidth. This is because gold has a higher density than silicon and adds more mass to the membrane. However, since the Young's modulus of gold is smaller than that of silicon, it produces a smaller effect on the stiffness of the membrane. This means that gold piston designs are easier to design for lower frequencies with smaller bandwidths and higher pressures.

The major limitation of the devices was found to be the voltage that could be applied to the device before the materials, especially the silicon oxide insulation layer, broke down. For a given gap height, the larger voltages that can be applied yield a larger electric field in the gap, which yield a larger pressure. We are currently fabricating these piston designs.

B. Transducer Design and Simulation

Using the CMUT membranes designed above, we developed an initial array for application below the ribs. This array is a 3.5 cm diameter 8-element, equal area, square phased array, shown in Fig. 6. While a circular array produces higher power gain, we chose this design for simplicity of the layout. This array was designed for application of therapy 3-5.5 cm beneath the skin surface with lesion size of 0.5 cm after several seconds.

1) *Simulation Methods:* We used Huygen's principle [20] to calculate the pressure profile. The pressure field was then converted to a power per unit volume using the following equation.

$$\langle q \rangle = 2\alpha I \quad (1)$$

where α is the absorption of liver tissue and I the intensity. The intensity is given by $P^2/(2Z)$ where P is the pressure at that point and Z is the impedance of the medium. Using the Pennes-Bioheat equation in a homogenous medium [22], these beam profiles were used to calculate temperature isocontours as a function of time. We used the common DSS044 and ODE routines in Matlab to solve the Bioheat PDE in Cartesian coordinates in three dimensions.

$$\rho c \frac{\delta T}{\delta t} = k \nabla^2 T + (\rho c)_b w_b (T_a - T) + q_m, \quad (2)$$

where T is the temperature, ρ is the density of the tissue, c is the speed of sound in tissue, k is the heat conductivity, ρ_b is the density of blood, w_b is the perfusion, c_b is the speed of sound in blood, T_a is the ambient temperature, and q_m is the heat input that that point. The resulting temperatures were then used to calculate the CEM_{43} .

$$CEM_{43} = \sum_{t=0}^{t=final} R^{(43-T(t))} \Delta t, \quad (3)$$

where $R = 0.25$ if $T(t) < 43$ and $R = 0.5$ if $T(t) > 43$.

2) *Simulation Results:* The pressure profile is shown in Fig. 7; the -6dB contour, is 2 mm x 2 mm x 7.5 mm. Assuming a homogeneous media with attenuation of 0.4dB/cm/MHz [21], a 1 MPa peak surface pressure will produce a 3.5 MPa peak pressure at the focus. A intensity of about 300 W/cm² will be deposited in the focal spot.

These preliminary results indicate the annular array will be able to form 0.5 cm lesions in liver tissue within minutes (Fig. 8); this a reasonable treatment time for liver ablation and the resulting lesion is a reasonable size for MR guidance. These results set the design requirements for the CMUT membranes. Since we must also account for cooling in large blood vessels and variability in perfusion, we will aim for a surface peak pressure of 1MPa.

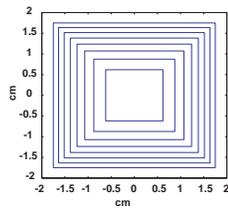


Fig. 6. Initial design for subcostal CMUT array. Concentric rings have equal area. When operated at 2MHz, the system will be able to focus at depths from 3-5.5 cm on axis.

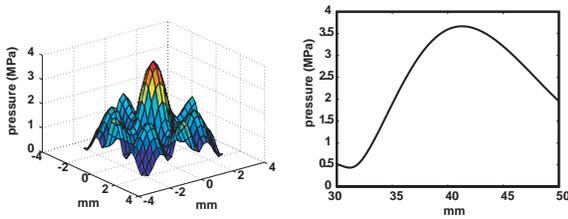


Fig. 7. Beam profiles calculated in the tissue when focusing at 4 cm. (left) The x-y beam profile at 4 cm demonstrates a beamwidth of 2 mm in the xy plane (as defined by -6db contour). (right) The z axis beam profile at x=0, y=0 demonstrates a lesion length of 7.5 mm in the xz plane.

IV. CONCLUSIONS

Typical imaging CMUTs have demonstrated capability for high pressure and continuous wave operation. Modifications of the CMUT structure to form piston-like membranes have been simulated and have been shown to output 1-2 MPa peak pressure needed for HIFU ablation. These membranes have been designed into a concentric, 8 element, equal area array that provides enough focal gain to produce 300 W/cm^2 at the focal point in liver tissue, accounting for attenuation and diffraction. With this power deposition, tissue could be ablated and the resultant temperature rises could be measured under MR guidance. Test transducers and annular arrays are currently being fabricated.

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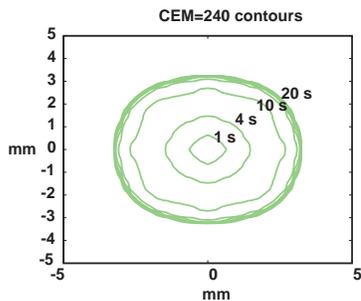


Fig. 8. Cumulative Equivalent Contours for 240 equivalent minutes, which indicates necrosis of liver tissues.

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