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MICROMACHINED CAPACITIVE ULTRASONIC IMMERSION TRANSDUCER FOR MEDICAL IMAGING

*X. C. Jin, B. T. Khuri-Yakub, F. L. Degertekin, I. Ladabaum, S. Calmes
Edward L. Ginzton Laboratory, Stanford University, Stanford, CA 94305
*Email: xcjin@alumni.stanford.org

Abstract

Piezoceramics have been the dominant transducer technology in ultrasound medical imaging for several decades. Recent progress in surface micromachined capacitive ultrasonic immersion transducers makes them an alternative transducer technology, especially in highly integrated two-dimensional arrays. This paper demonstrates that the surface micromachined capacitive ultrasonic immersion transducer performs at a level competitive enough to challenge the established piezoelectric transducers. Single element transducers and a variety of array transducers are fabricated with CMOS compatible micromachining technology. The transducers are observed to operate from 2MHz to 15MHz in immersion operation. Better than 100dB dynamic range is evident around 4.5MHz for a single device with only 6dB of unknown return loss. Similar performance is observed in a pulse echo experiment. The transducer's beam pattern indicates that the device behaves as a uniform piston transducer. Theoretical analysis shows that it is feasible to build an ideal immersion transducer with 100% of bandwidth and 3dB insertion loss in wide frequency ranges. The study in this paper concludes that micromachined ultrasonic transducers are an attractive alternative to piezoelectric transducers in ultrasound medical imaging.

Keywords

Medical ultrasound, ultrasonic transducer, micromachining, array imaging, capacitive transducer

I. Introduction

Ultrasound has been very successful as a medical imaging modality in the past decades. With its acceptance in many new medical applications, the requirement for the ultrasonic transducers becomes higher. Although piezoelectric ceramics and engineering cleverness have enabled a significant number of advanced ultrasonic devices and systems, many modern applications would benefit from transducers based on a different principle of actuation and detection.

As a result, active research work on capacitive micromachined ultrasonic transducers (cMUTs) has been reported recently. In addition to the wide operating temperature range (piezoelectrics depole at relatively low temperatures [1]), cMUTs have the potential to be integrated with electronics to form high density transducer arrays, and especially, immersion cMUTs are attractive for ultrasound medical imaging applications [2].

In our previous work, we demonstrated the successful operation of single element immersion cMUTs [3], [4], derived the electrical equivalent circuit model [5], [6], improved the microfabrication [7], [8] and discussed the optimization of fabrication process [9]. In this paper, systematic design, analysis, fabrication, and characterization of immersion cMUTs are discussed, especially targeting array transducers.

II. Transducer Fabrication

A cMUT consists of metalized silicon nitride membranes suspended above heavily doped silicon bulk. A schematic of one element of the device is shown in Figure 1 where t_n is the membrane thickness, t_a is the air gap thickness, t_b is the insulator layer thickness, r is the cell radius, sp is the cell support dimension, V_{DC} is the DC bias voltage, T is the tension within the membrane, and V_{AC} is the AC signal excited or received. A transducer consists of many such elements. When a voltage is placed between the metalized membrane and the bulk, Coulomb forces attract the membrane toward the bulk and stress within the membrane resists the attraction. If the membrane is driven by an alternating voltage, significant ultrasound generation results. Conversely, if the membrane is biased appropriately and subjected to ultrasonic waves, significant detection currents are generated.

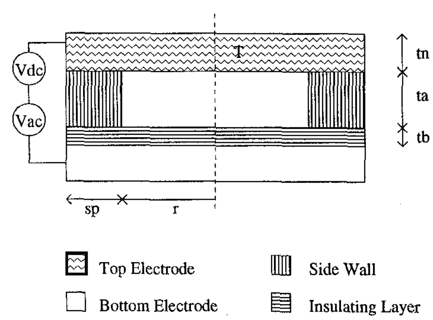

Fig. 1. An immersion cMUT element.

The fabrication scheme of the cMUTs used to generate the results herein reported is found in Figure 2.

The fabrication process starts with 4 inch n-type (100) silicon wafers heavily doped with 4 hour phosphorus gas phase drive-in at 1000°C to achieve good conductivity (better than 1.5Ω/□ resistivity) at the wafer surface. A thin layer of LPCVD nitride is deposited at 800°C as an insulator and an etch stop in the sacrificial etch to be performed later. A sacrificial polysilicon layer is subsequently

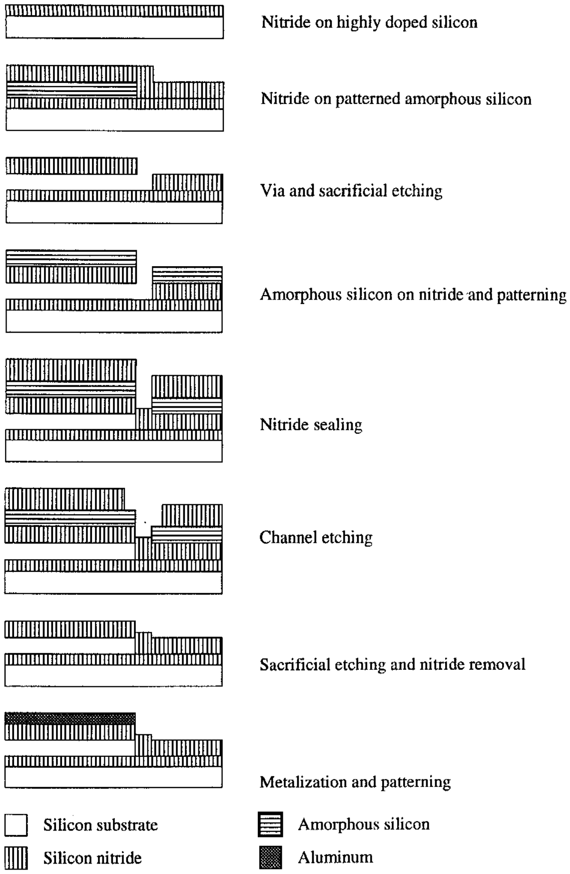

Fig. 2. Fabrication process of cMUTs.

deposited and dry-etch patterned into octagonally shaped islands to define the active transducer regions.

Nitride as the membrane material is then deposited with desired stress control. Vias are dry-etched to allow sacrificial etch, followed by vacuum sealing and selective removal of sealing material to preserve the original membrane thickness.

Lastly, aluminum is sputtered and wet etch patterned to act as the top electrode. The same aluminum deposition also defines bonding contacts to the bottom electrode through a lithographically defined trench in silicon bulk.

The top view SEM of such a device is shown in Fig. 3 with 5um spacing between two adjacent octagonal cavities.

III. Transducer Model

The basic cMUTs electrical equivalent circuit model is derived from the work of [10], [11]. The critical assumption is that the tension generated by a membrane displacement is small compared to the residual tension.

Approximating the cMUT as a parallel plate capacitor, its capacitance C is given by

$$C = \frac{\epsilon_0 \epsilon S}{\epsilon_0 t_n + \epsilon t_a} \tag{1}$$

where ϵ the dielectric constant of the membrane material,

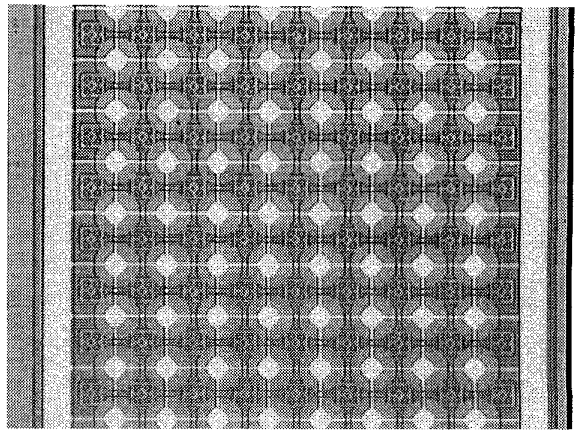

Fig. 3. SEM of a portion of an immersion cMUT .

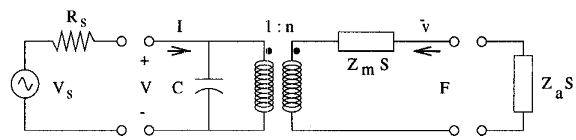

Fig. 4. Electrical equivalent circuit of cMUTs.

and S the area of the membrane. With the improved theory of cMUT as discussed in [3], [5], a transformer ratio

$$n = \frac{V_{DC} \epsilon_0 \epsilon^2 S}{(\epsilon_0 t_n + \epsilon t_a)^2} \tag{2}$$

can be derived so that we can write the current I as sum of electrical and mechanical components

$$I = C \frac{d}{dt} V - n\bar{v} \tag{3}$$

It is clear that n can be made larger by increasing the applied voltage or by decreasing the membrane and air gap thickness. Now we can draw the small signal equivalent circuit shown in Fig. 4 where Z_a and Z_m are the immersion load and membrane mechanical impedance, respectively, V_s and R_s are the electronic circuit parameters, F is the mechanical force, and v is the average membrane velocity.

The electrical equivalent circuit allows analysis and optimization of the cMUT structure. Parameters in the model can be varied to generate desired values in the equivalent circuit. Thus, it is clear that the structural control offered by microfabrication coupled with the insight from the equivalent circuit model can yield optimized transducers.

IV. Transducer Characterization

The transducers with the parameters shown in Table I are successfully fabricated. When a pair of 1.75mm x 1.75mm devices are used in immersion transmission, a signal to noise ratio of at least 48dB is obtained at the receiver. The return loss of the pair of immersion cMUTs is shown in Fig. 5. It includes (a) attenuation from impedance mismatch of both transmitter and receiver cMUTs with their

50Ω interfacing electronics; (b) loss from diffraction of the acoustic waves traveling from transmitter to receiver (2cm apart) in the liquid media; and (c) loss from all the remaining unknown sources. Taking into account the 20dB AC excitation headroom for the transmitter (to avoid saturation in electronics, 1v AC excitation is used), better than 100dB dynamic range is achieved around 4.5MHz for immersion cMUTs. It is noted that the immersion cMUTs have better than 100% bandwidth around its center frequency in Fig. 5. Similar transducer performance is observed in pulse echo experiment as shown in Fig. 6 with single burst excitation. It is evident that immersion cMUTs are very attractive alternatives to piezoelectric transducers in terms of performance of a single transducer.

Table I. Design Parameters for Immersion cMUTs.

Item	r	t_a	t_n	t_b	num
Experiment	15um	0.09um	0.53um	0.16um	2500

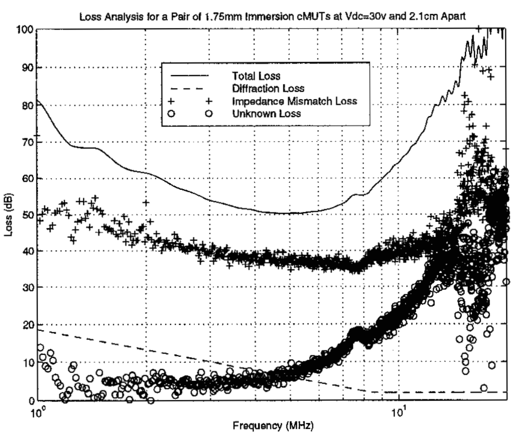

Fig. 5. Total return loss for 1.75mm device.

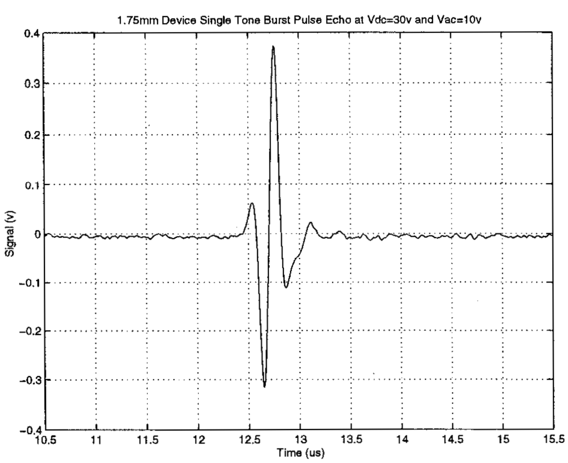

Fig. 6. Pulse echo response for 1.75mm device.

In order to characterize the feasibility of array applications of immersion cMUTs, the beam patterning experiment is also performed. When the pair of immersion cMUTs are placed at 8cm apart for transmission, a beam pattern plot as shown in Fig. 7 is obtained. The agreement between the experimental data and theoretical calculation [12] suggests that cMUTs behave like uniform piston transducers, and they are readily suitable for array operation.

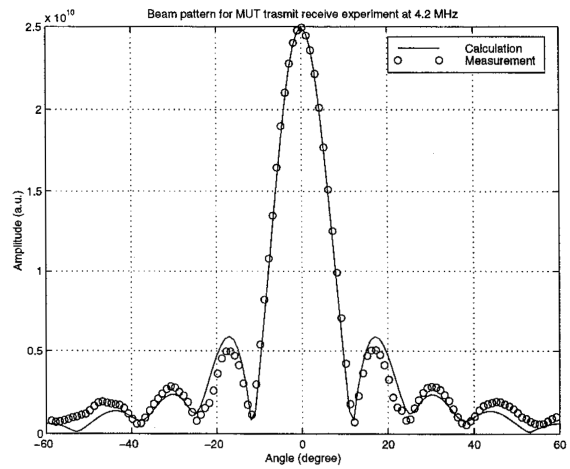

Fig. 7. 1.75mm immersion cMUTs beam pattern at 4.2MHz.

It is noted in Fig. 5 that the major loss is from the transducer impedance mismatch. Theoretical study based on the transducer electrical equivalent circuit model shows that it is possible to improve the transducer characteristics by placing the electrode underneath the membrane so that t_n is effectively reduced to zero in capacitor calculation. This ideal immersion cMUTs has less than 3dB of insertion loss (neglect diffraction loss) and 100% of bandwidth with almost equal real and imaginary part impedance of 50Ω around 4MHz, as shown in Fig. 8 and Fig. 9. Fabrication process to realize such kind of design is presently under investigation in our laboratory.

In order to test the array characteristics of the immersion cMUTs, a variety of structures are being built. Fig. 10 shows four 64-element linear arrays on a corner of a 4 inch wafer coated with photoresist. Each transducer element has 8 x 160 of octagonal cells as shown in Fig. 3. Ultrasonic array imaging is presently under investigation in our laboratory.

V. Conclusion

This paper presents an optimum immersion cMUT operating in the megahertz frequency range. The basic structure in microfabrication consists of silicon nitride membrane with a polysilicon sacrificial layer.

The immersion cMUTs fabricated with the process reported herein give a broad 50Ω real part impedance in the megahertz frequency range. When the transducer is untuned, it gives better than 100% of bandwidth with dynamic range of 100dB around 4.5MHz. Similar per-

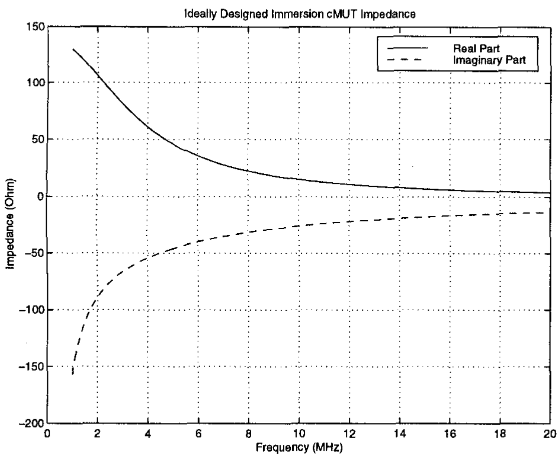

Fig. 8. Ideal immersion cMUT impedance.

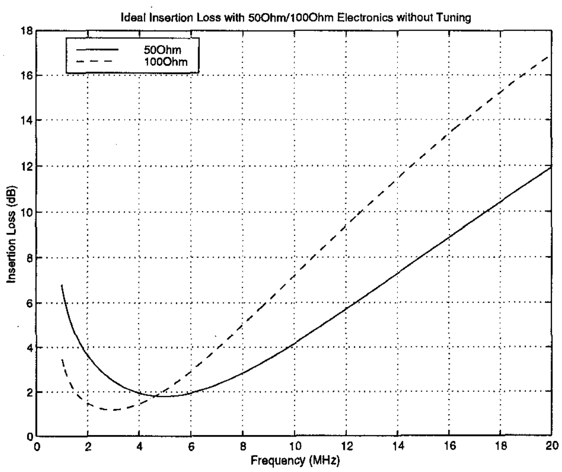

Fig. 9. Ideal immersion cMUT insertion loss.

formance is observed in pulse echo experiment. In addition, the beam pattern measurement shows the immersion cMUTs behave like uniform piston transducers and are readily suitable for array operations.

The results indicate that surface micromachined ultrasonic immersion transducers are an alternative to piezoelectric transducers in immersion applications.

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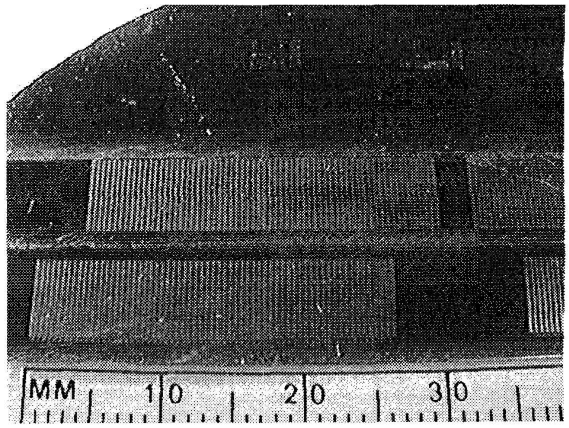

Fig. 10. 64-element linear array immersion cMUTs.

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