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An Improved CMUT Structure Enabling Release and Collapse of the Plate in the Same Tx/Rx Cycle for Dual-Frequency Acoustic Angiography

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Abstract

This study demonstrates in detail the potential of using capacitive micromachined ultrasonic transducers (CMUTs) for acoustic angiography of microvasculature. It is known that when

ultrasound contrast agents (microbubbles) are excited with moderate acoustic pressure around their resonance (2–4 MHz) they produce higher order harmonics (greater than 3rd harmonic) due to their nonlinear behavior. To date, the fundamental challenge has been the availability of a transducer that can generate transmit signals to excite the microbubbles at low frequencies, and in the same cycle confocally detect harmonics in the higher frequencies. We present a novel device structure and dual-mode operation of a CMUT that operates with a center frequency of 4.3-MHz and 150% bandwidth in conventional mode for transmitting and a center frequency of 9.8-MHz and a 125.5% bandwidth in collapse mode for receiving. An output pressure of 1.7 MPa_{pp} is achieved on the surface of a single unfocused transducer. The mechanical index at the transducer surface is 0.56. FEM simulations are performed first to show the functionality of the proposed device and then the device fabrication is described in detail. Finally, we experimentally demonstrate the ability to detect the microbubble signals with good contrast and the background reflection is adequately suppressed, indicating the feasibility of the presented approach for acoustic angiography.

Keywords

insulation spacers; microbubbles; low-frequency transmit; high-frequency receive; contrast-enhanced ultrasound imaging; dual-mode operation

I. INTRODUCTION

ULTRASOUND imaging techniques have made tremendous advancements over the last few decades and ultrasound is established as a non-invasive diagnostic imaging modality for clinical decision making [1]. Acoustic angiography is a high-resolution contrast-enhanced ultrasound imaging technique based on the superharmonic imaging of microbubbles injected into the blood stream [2], [3]. This technique is based on the ability of the microbubbles to oscillate when excited with an acoustic pressure (2 – 4 MHz) and subsequently produce a highly nonlinear acoustic response (from fundamental to greater than 3rd harmonic) [2], [4]. Since the tissue response is largely linear, bandpass filtering allows the separation of the nonlinear response of the microbubbles from the linear response of the surrounding tissue, resulting in little or no tissue background in the image [1], [2]. The detection of the higher-order harmonic signal generated by the microbubbles generates images with a high contrast-to-tissue ratio (CTR) and high-resolution [5]–[7]. Thus the ultrasound images produced with this technique illustrate microvasculature with high resolution and without the background noise from the surrounding tissue [3], [5]. The prevailing limitation of this imaging technique is that it requires transducers that can transmit a low-frequency (LF) pulse and also receive high-frequency (HF) harmonics, and this is a challenge for many conventional transducer arrays with narrow operating frequency range [7].

To date, several works have been reported to demonstrate the feasibility of superharmonic imaging [2], [4], [8]–[15]. Dual-frequency arrays containing two different types of elements arranged in an interleaved pattern (odd and even elements) to enable separation of transmission and reception modes [8], [16] have been shown, as well as dual-frequency confocal transducers with a low-frequency element for microbubble excitation and a second

element for the detection of the high-frequency content used by mechanical scanning [2], [3]. A 3 MHz/18 MHz co-linear array was used for superharmonic imaging [9], and a 2 MHz/14 MHz 1–3 piezo-composite transducer was employed in an in-vitro contrast ultrasound study using a tissue mimicking phantom [4]. A 1.7 MHz/20 MHz hybrid dual-frequency probe in a parallel geometry demonstrated an increase in CTR using superharmonic imaging relative to traditional high-frequency B-mode imaging [15]. However, there are difficulties and complexities, such as difficulty in coaligning high and low frequency elements and application of two different acoustic matching layers, that are associated with dual-frequency transducers with piezoelectric materials [17]–[19].

Capacitive micromachined ultrasonic transducers (CMUTs) are competitive candidates for medical ultrasound imaging due to their inherent advantages of wide operating frequency range, broad bandwidth, ease of array fabrication, and ease of integration with electronics [20]. Previously several multifrequency CMUT transducers for enabling advanced imaging and therapies have been explored. The frequency response of CMUTs is tunable by the bias voltage. Some works have explored the feasibility of switching between the two modes of CMUTs, conventional mode for a low-resolution deep-penetration to navigate, and collapse mode for high-resolution imaging and diagnosis [21]–[23]. It has been demonstrated that collapse-mode operation allows generation and detection of ultrasound more efficiently than the conventional mode of operation and additionally provide operating flexibility for dual-mode imaging systems [22], [24]. The usefulness of CMUTs for intracardiac echo (ICE) imaging has been explored. The center frequency of a CMUT transducer working in collapse mode was controlled between 8.7 MHz and 15.3 MHz by tuning the bias voltage [23]. A monolithic multiband CMUT array with multiple elements at different frequencies has been developed for mechanically scanned photoacoustic tomography [25]. A capacitive-piezoelectric hybrid ultrasound transducer consisting of a low-frequency (7 MHz) piezoelectric stack and a high-frequency (14 MHz) CMUT was developed for medical applications [26], [27]. A capacitive-piezoelectric hybrid structure which can operate at both CMUT and PMUT modes at both low frequency of 5.9 MHz and high frequency of 26.6 MHz, was developed for image-guided precision medicine [28]. A multi-frequency array was formed with interlaced large and small CMUT cells [29]–[31] for multi-band ultrasound imaging.

Earlier, we reported a high-frequency ultra-wideband (2–32 MHz) 1D CMUT array for acoustic angiography applications [32]. In this work, our proposed scheme is to transmit a low-frequency high-pressure pulse from a CMUT during release of the plate from collapse (snapback) and then switch to collapse-mode to receive the high-frequency microbubble harmonics in the same transmit-receive cycle. The transmit frequency for the excitation of the microbubbles can be tuned by controlling the time between releasing the plate and bringing it back to collapse. We accomplished this novel dual-mode operation of CMUT by improving our existing design [33] and creating insulating spacers in the device cavities. The insulating glass spacers inside the cavities prevent the direct contact of the top and bottom electrodes, and facilitate quick release of the plate and switching back to collapse-mode. Previously, we have presented preliminary results showing the operation of the proposed device with 4.8-MHz center frequency in conventional mode and 7.8-MHz center frequency

in collapse mode with a higher cutoff frequency of 13 MHz. We also showed that this structure helps to reduce dielectric charging [34].

In this paper, we present a detailed study of the feasibility for dual-frequency acoustic angiography using the proposed improved CMUT structure. In Section II, the FEM modeling of the device is described. The transmit and receive bandwidths of the dual-mode operation have been characterized in simulations and are later verified by the experimental characterization. Section III describes the complete fabrication process of the proposed devices. The device characterization both in air and immersion is presented in Section IV. The operation scheme of the device is also described in detail in this section. Preliminary imaging results are then presented in Section V which validate the feasibility of using the described mode of operation of the device for dual-mode acoustic angiography. Section VI discusses the limitations of the current setup and outlines further improvements.

II. DESIGN AND FINITE ELEMENT MODELING

The proposed device with spacers inside the cavity is simulated using a commercial finite element analysis (FEA) package (ANSYS v17, ANSYS Inc., Canonsburg, PA) as shown in Fig. 1. The spacers are arranged such that one is located at the center of the cavity and the rest form a hexagon around the central spacer. The CMUT plate is modeled using the SOLID185 element which has eight nodes where each node has three translational degrees of freedom in x,y, and z directions for each node. This is a suitable method for the 3D modeling of solid structures. The FLUID30 element is used for modeling the fluid medium to account for the acoustic loading and the interface fluid-structure interactions. The fluid structure interaction flag is enabled on the plate surface to add the fluid pressure load at the interface. To eliminate reflections in the fluid domain, a second-order boundary condition is realized using the FLUID130 element. The FLUID130 elements lie on a spherical boundary that completely enclose the domain meshed with FLUID30. The symmetric boundary conditions are applied on each side of the column to replicate the cell in x and y directions to effectively form a large CMUT element. The electrostatic gap is modeled using the transducer elements TRANS126. The TRANS126 element is a reduced-order model and couples the electromechanical domains between the conductive plate and the bottom electrode. The effect of the dielectric layer in the device is included by calculating the effective gap height. Two sets of TRANS126 elements are used to realize the spacers within the cavity. For the first set, the gap height is the distance between the top and bottom electrodes. For the second set, the gap height is the distance between the top electrode and the top of the spacers [Fig. 1(b)]. By adjusting the gap height of the TRANS126 elements, the effective locations of the spacers within the cavity are defined. There is no electrode under the spacers inside the cavity. The real constant FKN defines the normal contact stiffness factor and is set to 1 to overcome convergence problems with a reasonable penetration at the contact interface [24]. The TRANS126 elements are generated using the EMTGEN command. The device dimensions used in the simulations closely match the dimensions of the fabricated devices measured by using atomic force microscopy (AFM).

The static analysis shows the displacement profile of the circular plate under different DC bias conditions. The static simulation is performed in air with atmospheric pressure surface

load. The insulating glass spacers inside the cavities, prevent the direct contact of the top and bottom electrodes, even when biased at a voltage greater than the collapse voltage of the device. Fig. 2 shows the deflection of the plate in conventional mode, at the initial contact (touchdown), and when the plate rests further (collapse) on the spacers as the DC bias voltage is increased. We define touchdown as the point when the center of the plate is deflected and reaches the full gap distance gradually and just makes contact with the center spacer. This definition is to differentiate the behavior from collapse where the plate is suddenly pulled-in. On increasing the DC bias voltage beyond the touchdown voltage, the plate is pushed further and rests on the spacers forming an effective ring-shaped vibrating plate. The transient analysis begins with turning off the transient effects (TIMINT, OFF) in the first step to ensure that the membrane is stable under applied DC bias. Then the transient effects are turned on (TIMINT, ON) and the analysis runs for 1 μ s. Fig. 3 shows the frequency response of the seven spacer device in conventional and collapse modes of operation resulting from the Fourier transform of the transient response.

III. FABRICATION PROCESS OF THE CMUTS

We have previously established a fabrication process that combines the benefits of anodic bonding and patterned metal bottom electrodes on an insulating substrate [33], which can be realized using just three photolithographic masks. By adding just one more photolithographic step to this process, we accomplished the novel device structure described in this work. The device cavity is patterned to leave insulating glass spacers inside the cavity which act as posts within the cavity. In collapse mode these spacers support the top plate such that the plate rests on them forming effectively smaller cells with a higher resonant frequency within the larger cell cavity. Earlier, CMUTs with dielectric posts inside the cavity have been reported to improve device reliability [35]–[38]. Our spacers inside the cavity are formed by patterning the insulating glass substrate and do not have any electrode under the plate. Thus they enable the dual mode of operation without causing high electrical field in the dielectric layer. As a preliminary proof-of-concept fabrication run, we incorporated the additional process step to define the spacers inside the cavity in our previously existing single element device design.

A. Spacer Definition and Bottom Electrode Formation

We used a standard 0.7-mm thick, 100-mm diameter borosilicate glass wafer (Borofloat 33, Schott AG, Jena, Germany). The first step is to remove organic and other gross contaminants from the surface. The borosilicate glass substrate was cleaned using a standard piranha solution, which is made of a 3:1 mixture of concentrated sulfuric acid (H_2SO_4) with hydrogen peroxide (H_2O_2), for 15 minutes. In the first lithography step, we patterned and etched the device cavity [Fig. 4(a)]. The cavity pattern was defined using a 2- μ m thick negative photoresist (AZ-5214E IR, Clariant, Wiesbaden, Germany), which is well suited for lift-off. After patterning, the wafer was hard baked at 125°C for 5-minutes. This step allows for better adhesion of the photoresist to the substrate and also facilitates use of the photoresist as a mask for etching. We used reactive ion etching (RIE) with SF_6 gas to etch glass to create a 140-nm deep cavity. The gap height between the device plate and the top surface of the spacers is defined with the first etch step. With the next lithographic step, the

cavities are patterned and etched to define the regions for bottom metal deposition and also to form the insulating glass spacers in the cavities [Fig. 4(b)]. For the second etch step, a combination of dry and wet etch was performed. Dry etching is anisotropic and helps define and preserve the shape of the spacers in the cavity, each of which is 5 μm in diameter. This etch step was concluded with a short 2-minute 10:1 buffered oxide etch (BOE) step to form an undercut to make the lift-off process easier after metal evaporation for the bottom electrodes. The lateral etch rate for BOE etching has been measured to be 200 nm/min. The wet etch also allows for a uniform etching and minimal surface roughness inside the cavities. The cavities were further etched in the locations defined for bottom metal electrode. The etch depth in this step was 225 nm. A 20-nm chromium and 130-nm gold layer was deposited by e-beam evaporation into the cavities to form the bottom metal electrode [Fig. 4(c)]. In the final structure, the height of the spacers from the bottom metal electrode was 75 nm, and gap height from top of the spacers to the top plate is 140 nm. The gap height between the top and bottom electrodes is 215 nm. Fig. 5 shows the AFM image depicting the spacers formed inside the cavity.

B. Plate Formation Using Anodic Bonding

In the next step, an SOI wafer is bonded to the patterned glass using anodic bonding. The device layer of the SOI wafer forms the plate of the CMUT structure. Prior to bonding, a 200-nm silicon nitride layer was deposited on top of the device layer of the SOI wafer by using plasma enhanced chemical-vapor deposition (PECVD) [Fig. 4(d)]. The process conditions are discussed in detail in [33]. The high electrostatic field to which the glass substrate is exposed during the bonding causes outgassing in glass and trapped gas inside the cavities. To achieve a vacuum-sealed cavity, this trapped gas must be evacuated. Before the evacuation of the gas, the handle wafer of the SOI substrate was removed as in [33]. Different than our earlier approach, here we leave the buried oxide (BOX) layer to use it as a buffer layer in the subsequent silicon nitride etch step.

C. Evacuation of Gas and Formation of Electrical Contacts

In this step, we perform the evacuation of the trapped gas by creating an opening at the pad locations and simultaneously achieve access to the bottom electrode for forming the bond pads [Fig. 4(e)]. The silicon was etched with RIE using SF_6 gas. Then 2- μm PECVD silicon nitride was deposited to seal the cavities [Fig. 4(f)]. Reliability of sealing is critical for operation of CMUTs in immersion. The nitride thickness is chosen to be more than three times of gap height, to ensure successful sealing [39]. At this step, the cavities are vacuum-sealed. To achieve the final device structure and to access the bottom electrode, the silicon nitride layer is etched away, leaving it only at the sealing locations [Fig. 4(g)]. The BOX layer of the SOI wafer acts as a buffer in this step. Hence, the nitride can be slightly over-etched to make sure the metal pads are completely exposed, without over-etching the plate. This also ensures good uniformity of the plate. The BOX layer was removed using 10:1 BOE. The bottom metal pads were protected using photoresist prior to this etch step. Finally, a 20-nm chromium and 120-nm gold layer were deposited by e-beam evaporation and then lifted off in N-methyl-2-pyrrolidone (NMP) solvent to form the top electrode contact. This concludes the fabrication process and the final device structure [Fig. 4(h)].

IV. DEVICE CHARACTERIZATION

The dimensions of the final device structure are listed in Table I. Prior to microbubble detection test, the acoustic properties of a single element are characterized both in air and in immersion. The electrical impedance measurements in air demonstrate the functionality of the CMUT element both in conventional and collapse modes. The center frequency (f_c) and bandwidth are characterized. The output pressure and the mechanical index (MI) of the transducer are measured.

A. Acoustic Characterization in Air

The fabricated devices were first tested in air using a network analyzer (Model E5061B, Agilent Technologies, Inc., Santa Clara, CA) using a 250- μ m pitch coplanar microwave probe (Model ACP40-GS-250, Cascade Microtech, Beaverton, OR, USA) and an external high-voltage DC power supply (Model PS310, Stanford Research Systems, Inc., Sunnyvale, CA). The measured real and imaginary parts of the electrical input impedance in air for a seven-spacer device are shown for bias voltages of 65, 80, and 90 V in Fig. 6.

At 65 V, the CMUT is in conventional mode and the open-circuit resonant frequency of the device is 5.3 MHz. The touchdown voltage of the seven-spacer device is approximately 80 V. This was determined by observing the increase in the resonant frequency from 5.3 MHz to 10.1 MHz due to the center of the plate being in contact with the spacers, and effectively forming a ring-shaped vibrating plate. As the DC bias increases further, the plate is deflected more and rests on the spacers. At 90 V the plate appears to have an effective annular geometry with the center and sides clamped [40]. After contact with increasing DC bias the equivalent membrane size decreases, and hence the resonant frequency increases to 15.5-MHz. This means that the operating mode and the frequency response of the CMUT can be tuned by adjusting the DC bias.

B. Acoustic Characterization in Immersion

For tests in immersion, a small acrylic tank was built over the device that was wire bonded directly on a printed circuit board (PCB). Vegetable oil was used as the medium since the device surface was not electrically insulated. A calibrated hydrophone (Model HGL-0200, Onda Corporation, Sunnyvale, CA, USA) connected to a preamplifier (Model AH-2010, Onda Corporation) was placed at 9.3 mm distance from the transducer surface over the center of the transducer.

To analyze the frequency response in different operating modes, the element was biased at 65 V, 80 V, and 90 V, and excited with a unipolar pulse with an amplitude of 10 V and 25-ns pulse width. Fig. 7 shows the signal received by the hydrophone for the three cases. The Fourier transform of the received hydrophone signals was calculated. Fig. 8 shows the normalized frequency spectrum after compensating for frequency-dependent attenuation and diffraction losses and the pulse shape. The 3-dB fractional bandwidth is 150% in conventional mode (65 V) and 125.5% in collapse mode (90 V). The resonant frequencies in air and center frequencies in immersion as well as the 3-dB bandwidth in immersion at different DC bias voltages are summarized in Table II.

For the acoustic angiography application, the CMUT is biased such that it is initially in collapse-mode. The low frequency transmission signal for exciting the microbubbles is generated during the release of the plate from collapse and quickly switching it back to collapse in the same pulse-echo cycle. For receive, the device stays in collapse to detect the harmonics from the microbubbles. The microbubble excitation frequency generated can be tuned by controlling the time between releasing the plate and bringing it back to collapse (the pulse width of the CMUT excitation). Here, the transducer was biased at 90 V (collapse mode). A negative 90-V, 140-ns pulse was superimposed on the DC bias to release and collapse the plate for transmitting the low-frequency excitation signal. This generates a transmit pulse with a center frequency of 3.1 MHz and the 3-dB fractional bandwidth is 72%. The output pressure at a distance of 9.3 mm was measured using the hydrophone as 450 kPa_{pp} corresponding to an output pressure of 1.7 MPa_{pp} on the surface of the transducer (calculated by compensating for diffraction and attenuation losses) (Fig. 9). The peak negative pressure (PNP) on the surface is ~1 MPa corresponding to a mechanical index (MI) of 0.56 at the transducer surface.

V. MICROBUBBLE IMAGING

A seven-spacer single-element CMUT device was used for the imaging experiments. Microbubbles and saline solution flowing through a polyethylene (PE) tube is imaged. The contrast agent used in this work was an in-house formulation of phospholipid shell, C_4F_{10} core microbubbles that has been previously described [41]. The CMUT element was mechanically scanned over the tube to create a virtual CMUT array. The CMUT transmits an excitation signal by releasing and collapsing the plate and is kept in collapse for the receive in the same pulse-echo cycle.

A. Experimental Imaging Setup

The experimental setup for validation of the feasibility of dual-frequency acoustic angiography is shown in Fig. 10. A seven-spacer CMUT die was wire bonded directly on a PCB, which was attached to a 3D printed holder and mounted on a three-axis linear stage (Model PRO165, Aerotech Inc., Pitts-burgh, PA, USA) to allow mechanical scanning. A polyethylene tube (inner diameter of 580 μm and outer diameter of 965 μm) immersed in oil medium was used to flow microbubbles ($\sim 10^8$ microbubbles/mL). A syringe pump (Model 780210, KD Scientific, Holliston, MA, USA) was used to flow the microbubbles through the tube at a rate of 30 $\mu\text{L}/\text{min}$. The infusion rate is set such that fresh contrast is excited on each transmission. The same tube was used in two different scenarios. In one scenario, the tube was filled with microbubbles and in the other scenario, the tube was filled with saline solution. The CMUT was placed at a 4.8-mm distance from the tube. The peak-negative pressure at this location is 0.56 MPa corresponding to an MI of 0.3. The near field distance is 2.1 mm for 3.1 MHz transmit, which is important to make sure that we characterized the transmit performance of the single element in the far field. For imaging, the single element CMUT was mechanically scanned over the tube to create a 10-mm long virtual CMUT array. Synthetic aperture imaging approach was used for image reconstruction.

The data acquisition system, similar to [42], consists of a custom front-end analog circuit board for US transmission and reception; an analog-to-digital converter (ADC) board (PCI-5124, National Instruments, Austin, TX, USA) to digitize the received acoustic signals; an FPGA evaluation board (VC707, Virtex-7, Xilinx Inc., San Jose, CA, USA); a 3D linear stage; and a PC running a custom LabVIEW program to interface the digitizer and 3D linear stage.

B. Ultrasound Data Acquisition

In each round of ultrasound data acquisition, the PC sent a trigger signal to the FPGA so that it generated a unipolar excitation signal and sent it to the frontend analog circuit board. The high-voltage (HV) pulser (MAX14808, Maxim Integrated Inc., San Jose, CA, USA) converted the excitation signal to a unipolar HV signal with a peak value up to ± 100 V to excite the CMUT element. The CMUT was biased with a bias-T circuit to superimpose the HV transmit signal to the DC bias voltage. In receive, the acoustic signals were converted by the CMUT to electrical signals and amplified by a low-noise amplifier (MAX2034, Maxim Integrated Inc., San Jose, CA, USA). The signal went through a buffer (AD8044, Analog Devices Inc., Norwood, MA, USA) before being fed to the digitizer with a precision of 12-bits at a 200-MS/s sampling rate. The PC received the digitized acoustic data and notified the 3D linear stage so that it moved the CMUT element to its next position. The CMUT element was scanned for 10 mm in the y-direction with a step size of $50 \mu\text{m}$, synthesizing a virtual array of 201 elements. The imaging plane of this array was perpendicular to the tube (imaging target). In each transmit-receive cycle, the CMUT was biased in collapse mode (90-V DC) at the beginning. The CMUT was excited by a negative 90-V, 140-ns pulse superimposed with a 90-V bias, releasing the plate. This sent a 3.1-MHz excitation pulse (Fig. 9), and the reflections were received in the collapse mode. Saline solution and microbubbles were flowed sequentially through the tube.

C. Imaging Results

Ultrasound images were reconstructed with basic synthetic aperture imaging, which can be expressed as [43]:

$$I(\vec{r}) = \sum_{i=1}^N S_i(2 * |\vec{r} - \vec{r}_i|/c) \quad (1)$$

where $I(\vec{r})$ is the pixel intensity at position $I(\vec{r})$; $S_i(t)$ is the A-scan signal received by element i ; $\vec{r}_i$ is the position of element i ; and c is the speed of sound. Prior to ultrasound image reconstruction, the A-scan signals were processed with two different bandpass filters (2–5 MHz for low-frequency receive and 9–13 MHz for high-frequency receive). Then Hilbert transformation was applied which converted each A-scan sample from a real value to a complex one, where the real part was the in-phase component and the imaginary part was the quadrature. Therefore, each pixel of the image resulted from equation (1) would be a complex number. The real part forms an “I” image and the imaginary part forms a “Q” image. Envelope detection was applied by taking the absolute value of each pixel, and logarithmic compression was applied before displaying the image. We chose a “receive

high-frequency” bandpass filter around the microbubble harmonic frequency to be able to focus on the responses from microbubble harmonics and tune out responses from other frequency components, such as harmonics from the transmit signal itself. 9–13 MHz was chosen to maximize receive sensitivity (Fig. 8) and reduce harmonic content from the transmit signal (Fig. 9). This naturally results in a narrow-band filter for the high-frequency receive which limits the available resolution.

Fig. 11 shows the images for both the saline solution and microbubbles reconstructed using “receive low-frequency” (2–5 MHz) and “receive high-frequency” (9–13 MHz) filters. Fig. 12 shows the relative signal intensity (dB) of signal strength inside the tube relative to the reflection from the tube wall calculated for the four imaging cases.

In the case of transmit at low-frequency and receive at low-frequency, the tube walls are seen irrespective of what is inside the tube, saline solution [Fig. 11(a)] or microbubbles [Fig. 11(b)]. In this frequency range of the filter, harmonic information cannot be detected and there is no significant signal from inside the tube. The reflections from the wall of the tube dominate in both cases [Fig. 12(a) and (b)].

Fig. 11(c) shows the high-frequency receive for saline solution. Since no harmonics are generated from the saline solution, the content inside of the tube cannot be seen. However, the tube walls can be seen perhaps due to the presence of some non-linearity in the transmitted signal. In the high-frequency receive case, it is clearly demonstrated that microbubble harmonics are present and can be distinctly detected [Fig. 11(d)] with adequate suppression of the tube wall background. The signal from the microbubble harmonics is significantly stronger (>3 -dB) [Fig. 12(d)] than the harmonic content of the transmitted signal.

VI. DISCUSSION

This study primarily proves the feasibility of the proposed dual-mode operation and demonstrates its potential for acoustic angiography. With the improved device structure the CMUT can be operated by releasing the plate from collapse and bringing it back to collapse for transmitting a low-frequency pulse and then receiving in collapse-mode in the same transmit-receive cycle. Earlier we have demonstrated that the use of a glass substrate reduces the parasitic capacitance and improves dielectric reliability of the device [33]. In this work, we have enhanced the three-mask process for fabricating vacuum-sealed CMUTs on a glass substrate and patterned the cavity to have insulating glass spacers. These spacers act as posts inside the cavity which allow us to achieve the basic dual-mode operation proposed in this study. The insulating glass spacers inside the cavity also offer the advantage of alleviating any potential charging in the dielectric layer. In collapse mode, the top and bottom electrodes are not directly in contact through the dielectric layer and hence the buildup of charge due to contact is limited. In our previous work, we observed more repeatable behavior of these devices and its charging profile has been presented [34].

The presented device operates in conventional mode with a center frequency of 4.3 MHz with a 150% fractional bandwidth in immersion. In collapse mode it operates at 9.8-MHz

with 125.5% fractional bandwidth in immersion. With a single unfocused CMUT element we were able to generate an acoustic pressure of 1.7 MPa_{pp} (PNP of 1 MPa) on the surface which corresponds to an MI of 0.56. For preliminary demonstration a polyethylene tube was imaged first with saline solution and then with microbubbles flowing through it. As shown in the imaging results, the microbubbles are detected with good contrast, given the unaccounted nonlinearities from the CMUT and maybe from the oil because much better tissue suppression has been shown. suppressing the reflections from the tube wall.

One challenge posed by using CMUTs for harmonic imaging has been the limitation introduced due to the harmonic content of the transmitted signal being present in the received signal [44]. CMUTs are inherently non-linear because the electrostatic force depends on the square of the applied voltage and the membrane displacement [44]–[46]. Hence undesirable harmonic components are present in addition to the fundamental excitation, which can be predicted by modeling the CMUT behavior [47]. Traditional harmonic imaging with CMUTs faces this complication, and several different methods have been proposed demonstrating the opportunity of using CMUTs for this imaging modality [46], [48]–[51]. Some of the approaches to reduce or control the non-linear effects of CMUTs include reduction of specific harmonics by utilizing compensation methods to eliminate non-linear components [46], adding mass to the CMUT plate to control the harmonic vibrations of the CMUT [52] and, adding a series impedance to the CMUT [50]. A detailed analysis of the CMUT harmonics generated in dual-mode operation and methods of transmit-harmonic compensation are necessary future steps. This work will be extended further to full CMUT array design, and these devices will be characterized comprehensively and evaluated with tissue-mimicking phantoms for in-vitro study of acoustic angiography application using CMUTs.

VII. CONCLUSION

In this paper we present a novel mode of operation of CMUTs for acoustic angiography applications. The CMUTs presented here have insulating glass spacers in the cavities. These spacers act as post structures within the cavity that support the top plate in collapse mode. The spacers also reduce dielectric charging of the insulation layer by limiting the contact of the dielectric layer with the bottom electrode and hence improve device reliability. For the acoustic angiography application, the CMUT is operated by releasing the plate from collapse to generate a low frequency transmit pulse to excite the microbubbles and subsequently operated in collapse mode to receive the high-frequency harmonics from the microbubbles in the same transmit-receive cycle. First, finite element modeling is used to verify the operation of the proposed idea and the simulation results are found to be in good agreement with measurements. The feasibility of acoustic angiography has been demonstrated with the proposed device structure and mode of operation by showing clear high-frequency signal content generated by nonlinear oscillations of microbubbles with strong signal intensity and adequately suppressed background reflections from the surrounding. i.e. the tube walls. This work will be adopted and extended further into CMUT arrays, with optimized design to achieve better separation of the transmit and receive bandwidths, specifically designed for acoustic angiography applications.

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Biography

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Fig. 1.
Finite element model used for simulations (a) top view (b) cross-sectional view

Fig. 2. Simulated static deflection profile of the seven-spacer device at different DC biases, 65 V, 80 V and 90 V.

(a)

(b)

Fig. 3. Small signal characterization of the frequency spectrum of the seven-spacer device in simulation (a) conventional mode representing the transmit (b) collapse mode representing the receive mode of operation. Figure insets display simulated time responses.

Fig. 4.

(a) Defining cavities by first lithography step and forming the cavities by first etch step. (b) Defining the bottom metal regions and the spacer locations by the second lithography step. (c) Defining the spacers by second etch step and depositing metal to form bottom electrodes. (d) Forming the plate by bonding a SOI wafer on glass using anodic bonding. (e) Evacuating the trapped gas by etching silicon on the pad locations. (f) Sealing cavities in vacuum by PECVD nitride deposition. (g) Sealing nitride removal gives electrical access to bottom electrode. (h) Forming contact pads by BOX removal, metal deposition and liftoff and the final device structure. (The 3D depiction shows only one spacer inside the cavity for simplicity).

Fig. 5.
AFM image depicting cavity and spacers defined within the cavity and bottom metal deposited

Fig. 6. Input impedance measurements at different bias conditions (a) real and (b) imaginary parts.

Fig. 7. Small-signal impulse response and bandwidth (uncompensated) of a seven-spacer device measured at 9.3 mm for (a) 65-V (b) 80-V and (c) 90-V DC bias. Signals are received by a broadband hydrophone.

Fig. 8. Small-signal bandwidth characterization of the seven-spacer device excited by a 25-ns 10-V unipolar pulse at different bias voltages. Signals are received by a broadband hydrophone. Spectra shown after compensating for frequency-dependent attenuation and diffraction losses and the pulse shape.

Fig. 9.
The microbubble is excited with a 3.1 MHz pulse with a pressure of 1.7 MPa_{pp} on the surface.

Fig. 10. Diagram showing the imaging test setup. Saline solution and microbubbles were flowed through the tube sequentially during the experiment.

Fig. 11.
 (a) Saline solution “receive low” (2–5 MHz) (b) Microbubbles “receive low” (2–5 MHz) (c)
 Saline solution “receive high” (9–13 MHz) (d) Microbubbles “receive high” (9–13 MHz).
 The dynamic range of the images is 20 dB.

Fig. 12. Quantitative comparison of the relative signal intensity (dB), ratio of the mean signal strength inside the tube (μ_{Ic}) to the mean signal intensity of the PE tube wall reflection (μ_{Iw}), for the four imaging cases. Figure inset depicts a cross-section of the tube and the ROI for μ_{Ic} and μ_{Iw} .

TABLE I**PHYSICAL PARAMETERS OF THE CMUT ELEMENT**

Parameter	
Cell diameter, μm	60.4
Plate thickness, μm	1.5
Silicon nitride insulation thickness, μm	0.2
Gap height from plate to the spacer top, μm	0.14
Height of the spacers from bottom electrode, μm	0.075
Gap height between the top and bottom electrodes, μm	0.215
Width of the spacers, μm	4.2
Distance between the spacers, μm	5.8
Bottom metal thickness, μm	0.15
Diameter of the element, mm	2
Shape of the cells	circular
Shape of the element	circular
Number of cells per element	240

TABLE II

RESONANT FREQUENCY IN AIR, IN IMMERSION AND –3DB BANDWIDTH AT DIFFERENT DC BIAS CONDITIONS

DC bias	f_o in air (MHz)	Q-factor in air	f_c in immersion (MHz)	BW in immersion (MHz)	FBW(%)
65 V	5.3	97.5	4.3	1–7.5	150
80 V	10.1	78.6	4.9	1.1–8.7	155.1
90 V	15.5	47.6	9.8	3.7–16	125.5