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Analysis and Design of High Frequency 1-D CMUT Imaging Arrays in Non-collapsed Mode

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Abstract

High frequency ultrasound imaging arrays are important for a broad range of applications, from small animal imaging to photoacoustics. CMUT arrays are particularly attractive for these applications as low noise receiver electronics can be integrated for an overall improved performance. In this paper we present a comprehensive analysis of high frequency CMUT arrays based on an experimentally verified CMUT array simulation tool. The results obtained on an example, a 40 MHz 1-D CMUT array for intravascular imaging, are used to obtain key design insights and tradeoffs for receive only and pulse-echo imaging. For the receiver side, thermal mechanical current noise, plane wave pressure sensitivity, and pressure noise spectrum are extracted from simulations. Using these parameters, we find that the receiver performance of CMUT arrays can be close to an ideal piston, independent of gap thickness and applied DC bias, when coupled to low noise electronics with arrays utilizing smaller membranes performing better. For pulse-echo imaging, thermal mechanical current noise limited SNR is observed to be dependent on the maximum available voltage and gap thickness. In terms of bandwidth, we find that Bragg resonance of the array, related to the fill factor, is a significant determinant of the high frequency limit and the fluid loaded single membrane resonance determines the lower limit. Based on these results, we present design guidelines requiring only fluid loaded single membrane simulations and membrane pitch to achieve a desired pulse-echo response. We also provide a design example and discuss limitations of the approach.

Keywords

High frequency arrays; CMUT; CMUT design; Phased array; Bragg's resonance; Photoacoustic receiver

I. Introduction

High frequency ultrasound imaging has been of an increasing interest for a broad range of applications. Ultrasonic devices with operation frequencies above 30 MHz can provide microscopic spatial resolution making these devices especially ideal for applications such as intravascular blood vessel imaging (IVUS) [1–3], small animal imaging [4–6], breast biopsy procedures [7], photoacoustic imaging [8–10], ophthalmology [11–13], and dermatology [14–16].

High frequency ultrasound images can be obtained by either mechanically scanning a single ultrasonic element, or by electronically scanning ultrasonic elements arranged in an array. Array based systems do not require any mechanical motion, enabling increased frame rate,

reliability, and resolution. In addition to these benefits, array-based systems allow imaging methods unattainable with a single element system such as dynamic focusing, aperture apodization, and electronic beam steering, making them desirable for high frequency ultrasound imaging [17].

An ideal high frequency phased array would require an element pitch close to half of the ultrasound wavelength to avoid grating lobes. Fabrication of transducer elements with such small element size is challenging with conventional piezoelectric dice-and-fill methods. Piezoelectric arrays that overcome this limitation employ different fabrication methods such as DRIE etching [18–20], partial dicing [21], laser micromachining [22], and kerfless arrays [23]. In comparison, conventional fabrication methods of capacitive micromachined ultrasonic transducers (CMUTs) can deliver features of several micron size with high yield [24]. As a result, CMUT fabrication technology can produce arrays with variable array geometries and small pitch sizes with relative ease. These fabrication methods also enable electronics integration, reducing the device size and parasitic capacitances, whereas special consideration is required for integration of piezoelectric arrays. Such an integration is especially crucial for devices with limited dimensions, such as IVUS systems potentially integrated on a guidewire [24, 25].

There has been significant work on high frequency CMUT transducer arrays in literature, with operation frequencies ranging from 20 MHz up to 50 MHz [26]. Fabricated arrays utilized diverse membrane geometries, ranging from 20 μm squares [24] to circles of varying diameters [27–30], to construct equivalently diverse array geometries supporting element pitches smaller or in the order of the ultrasound wavelength, ranging from 25 μm [24] to 65 μm [27]. These arrays exhibited a large variation of -6 dB fractional bandwidth, ranging between 28% [24] up to 100% [27]. However, measurement of these arrays were conducted in a variety of media with different attenuation characteristics which can significantly skew the results [26]. Despite the wealth of experimental data on these CMUTs operated in the non-collapsed mode, detailed modeling results have been lacking.

Several approaches have been introduced to model CMUT arrays operating in the non-collapsed mode [31–35]. A model based on small-signal equivalent circuit of an individual circular CMUT membrane was developed and used to evaluate crosstalk waves on CMUT arrays, but the model was limited to single vibration mode at low frequencies [36]. A more thorough analysis of a single element CMUT composed of an array of membranes was carried out utilizing boundary element method (BEM) with normal mode theory [37]. Modal decomposition of the admittance matrix was carried out to determine the modes of the element. The effect of CMUT layout and mechanical impedance on these modes were analyzed to obtain design guidelines to optimize the frequency response of the element in the transmit mode. That study however considered only a single element without considering the neighboring elements in an imaging array and the finite size in one of the dimensions, limiting its applicability to realistic imaging array design. Therefore, there is a need for full performance evaluation of the elements in an imaging array with signal to noise ratio (SNR) evaluation in the pulse-echo mode and receive only applications such as photoacoustic imaging [38] and passive imaging [39]. This type of comprehensive analysis requires a model that incorporates nonlinear CMUT behavior during transmit operation, along with any

transmit and receive electronics that might be integrated to high frequency CMUT arrays. The experimentally verified model developed by Satir *et al*/has demonstrated capability of simulating large-signal dynamics of CMUTs [40] and integrated electronics [41] which makes it suitable for this study.

In this paper, we first briefly describe the model utilized in our analyses and how relevant parameters, such as input impedance, transmit and receive sensitivity, are obtained for arbitrary array structures. We use the radiation impedance of an ideal rectangular array element as a benchmark for receive noise performance. We then focus on a 1-D phased array with 40 MHz center frequency as a case study. We evaluate thermal mechanical noise limited, pulse-echo based SNR for various designs considering the effect of different design parameters, such as the membrane geometry, DC bias, gap thickness, membrane and array pitch, and operation medium. The results are then analyzed to elucidate important design guidelines for optimized array response and a design example is provided.

II. Methodology

A. Description of Model

The model used for the calculations in this paper [41] is comprised of three main sections visualized as blocks in Fig. 1. Block A calculates the electrostatic force on each electrode and accounts for the nonlinearity observed in large-signal CMUT modeling. Increased accuracy may be obtained by considering the modal deflections of the electrodes and dividing the electrodes into patches accordingly [40]. In the analysis, electrode patches are chosen to capture the 1st and 2nd symmetric modes, with the 2nd symmetric mode corresponding to the 6th vibration mode of the CMUT membrane. Block B evaluates the vibroacoustic behavior of the CMUT array as a linear multiple input multiple output (MIMO) system incorporating forces on the electrodes to average displacements. BEM is utilized to obtain finite impulse response (FIR) filters for each electrode patch that encompass the linear mechanical behavior including acoustic crosstalk due to fluid coupling and higher order membrane vibration modes. It should be noted that this model does not consider mechanical coupling between CMUT membranes through the substrate, such as Lamb waves in the silicon wafer. These two blocks define the electromechanical behavior of the CMUT array. Block C is a multiple input single output (MISO) system that evaluates pressure at the desired point in the immersion fluid from the forces acting on the electrode patches. The effect of flat reflectors can be incorporated into the model by modifying the Green's function utilized in block B and C. Further analysis can be conducted by considering the CMUT array as a linear time-varying capacitor in the electrical domain. This forms a two-port network with four degrees of freedom for each element: input electrical current I_p , applied electrical voltage V_p , mean membrane velocity v_p , and external acoustic force F_p , enabling extraction of equivalent circuit parameters in receive mode, as well as integration of transmit/receive electronics.

B. Relevant Transfer Functions

Transfer functions that illustrate CMUT array element performance can be extracted by analyzing the relationship of the four variables of the two-port network in the model. The

model is first used to obtain small-signal equivalent circuit parameters for receive mode operation [42]. The electrical impedance at a certain frequency is calculated by dividing the input voltage with the output current at that frequency while the acoustic force is kept at zero. A Gaussian pulse is used as input to obtain the electrical impedance across the frequency range of the Gaussian pulse. The calculated impedance is then utilized to obtain the thermal mechanical current noise of the CMUT array element [43]. Similarly, frequency dependent pressure sensitivity for plane wave incidence is extracted by dividing the spectrum of an applied Gaussian force pulse to the spectrum of the output current while the input voltage is kept constant, and the electromechanical transformer ratio is obtained by dividing the spectrum of the resultant mean velocity and current. Calculated thermal mechanical current noise and plane wave pressure sensitivity is then further combined to obtain a pressure noise spectrum defining the minimum detectable pressure of the CMUT array element. Each of these parameters is important for the receiver performance of different CMUT array designs.

Transmit analysis of a CMUT array can be carried out through analysis of the pressure output. However, a more thorough analysis would be that of the pulse-echo response. The output current of the pulse-echo response is determined by both transmit and receive characteristics. In the analysis, a plane reflector at a set distance is incorporated into the model, and the output current due to the echo of transmit pulses is calculated. Assuming low noise receive circuitry [44], a SNR based on the reflection current and thermal mechanical current noise is utilized as the basis of pulse-echo performance analysis of different designs.

III. Analysis

To illustrate the analysis methods, a 1-D phased array with 40 MHz center frequency operating in water is chosen as a case study. The motivating application for this case study is guidewire based IVUS imaging [24]. As shown in Fig. 2, the total array size is chosen to be $300\text{ }\mu\text{m} \times 500\text{ }\mu\text{m}$ for proper integration to a 0.014" guidewire. The maximum element pitch (d_e) is chosen to be 33% larger than the half wavelength of the operation frequency, resulting in a maximum pitch of $25\text{ }\mu\text{m}$ and a minimum of twelve elements. Each element is populated with square membranes comprised of silicon nitride (Young's modulus = 110 GPa, density = 2040 kg/m^3 with varying lateral dimensions (a), elevation (d_x) and azimuthal (d_y) membrane pitches, membrane thicknesses (h), and gap thicknesses (g). The stress in the nitride membrane is not considered. The electrode coverage of the membranes is kept constant at 56% and the isolation material is chosen as HfO_2 (relative permittivity = 15) of a thickness so that the electric field under operation is at 75% of the breakdown field strength [45]. Initial receiver and pulse-echo analysis is carried out on arrays with varying lateral dimensions and constant fill factor, followed by analysis on arrays with constant lateral dimensions and varying fill factor.

A. Receiver Analysis

Although CMUT arrays are capable of operating in transmit-receive mode, they are also favorable for applications that do not require CMUTs operating as transmitters.

Photoacoustic imaging [38] and passive imaging [39] have been conducted with CMUT

arrays operating in only receive mode. Therefore, to discuss important factors in CMUT performance an analysis focusing on the receiver performance of the CMUT arrays is conducted.

Three high frequency 1-D CMUT array geometries are considered. These arrays are populated with membranes of three different lateral dimensions: 10, 15, and 20 μm . The elevation (d_x) and azimuthal (d_y) membrane pitch are kept at the same value (d) and adjusted to obtain a fill factor (FF) of 64%. The membrane thickness (h) is adjusted to obtain a center frequency of 40 MHz. A comparison of the array geometries can be seen in Table 1. The receiver characteristic parameters are extracted for a DC bias of 60 V and the gap thicknesses (g) are adjusted for operation at 90% collapse voltage. Additional investigation on the effect of gap thickness and DC bias can be found in the Appendix.

a) Frequency Response: Since there are important differences between single element (SE) and imaging array element (AE) receiver characteristics, these cases are analyzed separately. The calculated thermal mechanical current noise spectra (based on electrical input impedance) for the CMUT membranes of Table 1 are plotted in Fig. 3, to point out salient features of these receivers dependent on lateral dimensions. In the frequency range above the center frequency, three different cut-off frequencies are observed for each geometry (reference dips 1, 2, and 3) for both single and array elements. These cut-off frequencies are due to the Bragg's resonance of the elements. Bragg's resonance arises as a result of scattering caused by the periodic discontinuity of acoustic propagation between membranes. Destructive interference of the scattered waves forms a band gap at the frequency c_0/d , where c_0 is the speed of sound of the medium [46]. This resonance acts as a limiting factor to the frequency response of the CMUTs, both in single element and imaging array element, with smaller pitches resulting in higher limits.

In the frequency range below the center frequency, local sharp peaks and dips are observed especially for the imaging array elements (reference peaks 4, 5, and 6). These extrema are caused by the crosstalk between elements of the imaging array. Crosstalk excites subsonic and supersonic array modes in a frequency region around the first vibration mode resonance of the fluid loaded single membrane, resulting in multiple resonant peaks and dips and effectively forming a lower frequency limit [36, 47]. These resonant peaks and dips are observed in arrays designed with pitches close to and below half wavelength of the operation frequency, with the effect enhanced as the array pitch decreases [48].

Crosstalk in the imaging array can also actuate higher vibration modes of single membranes (specifically the degenerate 2nd and 3rd modes which are the 1st asymmetrical modes of the square membranes shown as inset in Fig. 3), resulting in additional resonant peaks and dips around the frequency range of these vibration modes. The location of these peaks relative to the Bragg resonance depends on the thickness of the membrane. For example, these "array excited" asymmetric vibration modes manifest themselves as a peak in the imaging array element populated with 10 μm membranes (reference peak 7). For the imaging array element with 15 μm membranes, these modes overlap with the Bragg's resonance cut-off frequency, hence they are suppressed. Finally, for the imaging array element with 20 μm membranes, these resonances are above 125 MHz and hence they do not impact the response in the

bandwidth shown in Fig. 3. Note that these modes are “array excited” and require large number of membranes in the lateral dimension and phased excitation of the array element. Their effects are not pronounced in single element CMUTs considered here as these elements are only one or two membranes wide, and also when the whole imaging array is excited in phase (see for example Figs. 9, 12 in [49]).

Overall, one observes a spurious resonance free range of frequencies between the first mode of the single CMUT membrane on the lower end and either Bragg or a higher order membrane mode resonance on the higher end [37, 50]. The effect of Bragg’s resonance and crosstalk is also observed in pressure sensitivity. However, crosstalk is less prominent in pressure sensitivity, as that parameter is dominated by the in-phase excitation response of all membranes to a normally incident plane wave [49].

b) Pressure Noise: The most important measure of receiver performance is the pressure noise spectrum which defines the minimum detectable pressure. The pressure noise spectrum of a CMUT element is obtained by combining the noise current and pressure sensitivity as described above. As the ideal case, the pressure noise spectrum of a baffled rectangular piston with the same geometry as the single element of the imaging array is considered. The piston pressure noise spectrum is obtained from its radiation impedance [51], enabling comparison of the receiver performances. The pressure noise spectrum of single elements and imaging array elements, along with the pressure noise spectrum of the ideal piston of the two element sizes are presented in Fig 4. It can be seen that utilizing membranes with smaller lateral dimensions approaches the receiver performance to that of the ideal piston, improving both sensitivity and bandwidth. Bragg’s resonance manifests itself as an edge for single elements and a peak for imaging array elements. Crosstalk from neighboring elements increases the current noise and decreases sensitivity around the frequency region of single membrane resonance, resulting in several local maxima in pressure noise spectrum [47]. Similarly, in the array with 10 μm membrane geometry, the effect of crosstalk induced higher order membrane modes manifest themselves as another peak. These features are highlighted using the same numbering scheme of Fig. 3 for clarity. Overall, compared to single element CMUTs, imaging array CMUTs have lower minimum detectable pressure in the spurious free frequency range, due to higher pressure sensitivity to incident plane waves. This increase is a result of the effective boundary conditions differing from an infinite baffle since imaging array elements are surrounded by moving membranes. In addition, CMUTs with smaller membranes exhibit a pressure noise spectrum closer to the ideal piston over a broad frequency range (35–85 MHz for 10 μm membrane geometry).

B. Pulse-Echo Analysis

A complete performance analysis of a CMUT imaging array requires a transmit/receive analysis. In the following, pulse-echo response of imaging array elements of the previous geometries is analyzed under unipolar pulsing. The performance is quantified using thermal mechanical noise limited pulse-echo SNR as a basis.

For best receiver sensitivity, the CMUT elements are biased close to collapse during transmit. Considering this limitation, two transmit pulsing methods are feasible for pulse-

echo analysis: unipolar and bipolar pulsing. In unipolar pulsing, the DC Bias of the CMUT element is decreased to zero for the length of the pulse width, whereas in bipolar pulsing, the DC bias is first increased to a certain value and then decreased to zero for the length of the pulse width. As a result, bipolar pulsing can utilize the gap thickness more efficiently. It has been shown that bipolar pulsing can enable an increase of 6 dB SNR compared to unipolar pulsing [52]. However, bipolar pulsing requires more complex electronics and a larger chip area for each element. Therefore, unipolar pulsing is considered in the pulse-echo analysis.

The maximum transmitted pressure output of a CMUT element operated in the non-collapsed mode is proportional to the gap thickness. Considering that the thermal mechanical current noise limited receiver performance of a CMUT array is independent of the gap thickness (shown in Appendix), the gap thickness must be maximized for optimum performance. The limiting factor therefore becomes the maximum available DC bias defined by the electronics integrated to the CMUT element. For the case considered, 60 V is chosen as the voltage limit and the gap thickness is adjusted for operation at 90% collapse voltage. Additional analysis on the effect of operation voltage can be found in the Appendix.

The pulse-echo analysis is carried out with single unipolar pulse of 10 ns width, corresponding to $1/(2.5 \cdot f_{center})$ for a 40 MHz transducer which maximized the pulse-echo signal. The current signal due to the echo from a plane reflector positioned 5 mm from the CMUT array is analyzed (denoted as reflection current) to ensure far field operation. Comparison of the obtained reflection currents can be seen in Fig 5. Similar to the receiver analysis, an increase in bandwidth is observed when smaller membranes are utilized. It is also observed that the signal levels are higher for wider array elements (25 μm width for 10 μm and 20 μm membrane elements, 18.75 μm width for 15 μm membrane elements). The noise value for SNR calculation is obtained by integrating the thermal mechanical noise over 20–60 MHz frequency range, assuming a 100% fractional bandwidth around 40 MHz for the receive system. SNR is then calculated as the ratio between the peak to peak value of the reflection current and root mean square noise. The SNR results for the three different designs are summarized in Table 2. Analyzing the results, it is observed that utilizing larger elements improves the overall SNR, and utilizing smaller membranes improve both the overall SNR and the –6 dB fractional bandwidth.

The directivity of each imaging array element is examined through their transmit pressure beam patterns. Normalized beam patterns are shown in Fig. 6. It is observed that the 25 μm wide element comprised of 10 μm membranes exhibits the lowest –3 dB span at $\pm 31.5^\circ$, indicating a larger equivalent rectangular element width (19 μm) as compared to the 18.75 μm wide element comprised of 15 μm membranes and 25 μm wide element comprised of 20 μm membranes which exhibit 3dB spans of $\pm 50.6^\circ$ and $\pm 54.9^\circ$. However, the 10 μm geometry would provide a similar SNR over the $\pm 45^\circ$ range due its higher absolute SNR (Table 2). Overall, all three arrays would have radiation patterns suitable for phased array imaging.

Since the assumed application of the arrays is in intravascular imaging, the performance of the different array geometries in blood is also analyzed. In comparison to water, blood has similar density and speed of sound, but significantly higher attenuation [53]. As a result,

operating in blood does not alter the receiver performance significantly, but reduces the center frequency and SNR of the pulse-echo response. As the effect of attenuation increases with frequency, the center frequency shift is greater for arrays with higher fractional bandwidth. The characteristics of the reflection currents of the same setup operating in blood and the obtained pulse-echo SNR values can be found in Table 3. In all array geometries, operating in blood reduces the center frequency but increases the -6 dB bandwidth, which is in line with large fractional bandwidths observed in the literature when measurements are conducted in oil. In each case, the array element provides SNR value above 50 dB. Assuming tissue walls to be weak reflectors of -40 dB, it can be deduced that each array has the capability of imaging a tissue wall at 5 mm distance in blood.

C. Fill Factor Analysis

The previous analysis has demonstrated that utilizing smaller membranes improve the frequency response and SNR when the fill factor is kept constant. Further analysis considering the fill factor is carried out with the same process since this is a factor easily controlled during CMUT fabrication. Three 1-D CMUT arrays comprised of $10\text{ }\mu\text{m}$ membranes with different fill factors (F) are investigated. The azimuthal pitch (d_y) is kept constant while the elevation pitch (d_x) is varied to obtain fill factors of 64, 73 and 80 %. A comparison of the array geometries can be seen in Table 4.

As in the previous analysis, the receiver characteristic parameters of single elements and imaging array elements are extracted at a DC bias of 60 V and 90% collapse voltage operation. Unipolar pulsing with 60 V pulse voltage and thermal mechanical noise limited SNR of reflection current is chosen for the pulse-echo analysis. Noise is integrated over 2060 MHz and reflection current of a perfect reflector at 5 mm distance is simulated.

c) Receiver Analysis: The calculated plane wave pressure sensitivity of the elements is shown in Fig. 7. When d_x and d_y of these arrays are different, two different Bragg's resonances appear as two different cut-off frequencies (reference dips 1, 2, and 3). The two Bragg's resonances together determine the upper frequency limit, with the element geometry supporting the smallest d_x displaying the largest bandwidth. As the single membrane resonance frequency is identical, the behavior for frequencies below the center frequency is similar, with the lower frequency limit located around the single membrane resonance (reference dip 4). The fill factor mainly affects the frequency span above the center frequency resulting in higher bandwidth and center frequency with higher fill factors. The same trend is also observed in thermal mechanical current noise and pressure noise spectrum. Therefore, the improvement of bandwidth due to increased fill factor is mainly due to higher effective Bragg resonance.

d) Pulse-Echo Analysis: The reflection currents of the imaging array elements for 10ns pulse width can be seen in Fig 8. Similar to the receiver analysis, an increase in bandwidth and center frequency is observed with higher fill factor. Moreover, this increase is accompanied with increased signal strength since the active area also increases. The characteristics of the reflection currents and the obtained pulse-echo SNR values can be found in Table 5.

A similar analysis, in which the elevation pitch (d_x) is kept constant while the azimuthal pitch (d_y) is varied, yields similar results. An increase in fill factor results in an increase in bandwidth. However, as the element pitch (d_{el}) is defined by the azimuthal pitch (d_y), increasing the fill factor results in smaller element size, which in turn decreases the SNR values. Nevertheless, signal strength per area increase, and an increase in fill factor is accompanied by an increase in area normalized SNR.

IV. Design strategy

The comprehensive analysis of various design parameters and their effect on performance conducted in the previous section makes it possible to prioritize these parameters for a design process. In this section we propose simple design guidelines for high frequency 1-D CMUT arrays to obtain a desired pulse-echo frequency response and associated maximum SNR. The design process considers thermal mechanical current noise limited operation, i.e. the integrated transmit/receive electronics generates less or equivalent noise as compared to the CMUT array element.

The preceding receiver performance and pulse-echo analysis demonstrated that high fractional bandwidth (FBW) operation with high SNR is feasible for high frequency CMUT arrays with large fill factors and small membrane dimensions. The analysis determined two limiting factors that dominate the frequency response of high frequency CMUT arrays: the Bragg's resonance (f_{Bragg}) as the upper frequency limit, and the single membrane resonance (f_{single}) as the lower frequency limit. Therefore, for large operation bandwidth these two limits need to be sufficiently separated.

Among the two limiting factors, f_{Bragg} is dependent only on the membrane pitch and the medium, defined as $f_{Bragg} = c_0/d$, whereas f_{single} is dependent on more parameters. Therefore, f_{Bragg} should be determined first. For the broadest frequency response, the array area needs to be populated with membranes at the smallest pitch and maximum fill factor with the limiting factors determined by the fabrication methods. The upper frequency limit being set, the lower frequency limit can then be defined to obtain the center frequency (f_{center}) and bandwidth by adjusting the membrane thickness. Utilizing larger membrane thicknesses would result in higher f_{single} values, reducing the effect of crosstalk and increasing sensitivity at the cost of reduced bandwidth. As a result, a small membrane thickness value needs to be chosen for a low f_{single} value and large bandwidth. However, as the frequency span between f_{single} and f_{Bragg} increases, the upper frequency limiting factor will shift from f_{Bragg} to the 6th vibration mode (2nd symmetric mode) resonance frequency of the fluid loaded square CMUT membrane, denoted as f_{ares} . As a symmetric mode, f_{ares} affects far field operation more significantly than the lower frequency asymmetric modes. Similar to f_{single} , this frequency can be found from the single membrane response in fluid.

To evaluate a larger design space, additional simulations were conducted with membrane dimensions ranging from 7.5 to 20 μm and different fill factors (FF). The characteristics of the additional array geometries and their pulse-echo responses can be found in the inset table of Fig. 9. When the BW vs f_{Bragg_ave} is plotted, as in Fig. 9, a linear relationship in the form of $BW \sim f_{Bragg_ave}/4$ is observed, where average Bragg's resonance frequency is defined as

$f_{Bragg_ave} = 0.5 \cdot (c_0/d_x + c_0/d_y)$. The reason for linear dependence can be seen in Fig. 7. When f_{single} is fixed to a constant value determining the lower frequency limit, the BW is determined by slope of the frequency response above the center frequency which is in turn determined by f_{Bragg_ave} . Note that this BW estimation loses accuracy as f_{Bragg_ave} approaches f_{center} , where the Bragg resonance and single membrane resonance effects cannot be isolated.

Taking the frequency limits of the CMUT array and their mechanisms into consideration, an algorithm can now be proposed to determine the lateral dimensions and membrane thickness for a desired f_{center} and FBW . This is depicted in Fig. 10. As the first step, the frequency limits (f_{low} , f_{high}) are determined for a 1-D CMUT array operating at a center frequency of f_{center} and a fractional bandwidth of FBW . For large FBW and f_{center} limited operation, f_{Bragg} can then be determined from the frequency span. f_{Bragg} and f_{center} are then used to determine the membrane and element pitch (d , d_e). At this point, it is important to check if the calculated f_{Bragg} is larger than f_{high} . If $f_{high} < f_{low}$, the pitch has to be decreased to ensure the validity of f_{high} . If d and f_{Bragg} are reasonable, membrane lateral dimensions maximizing the fill factor are chosen. The thickness of the membrane (h) is then set to obtain a f_{single} close to f_{low} . While setting the thickness, one needs to check that $f_{ares} > f_{Bragg}$ holds to make sure that f_{Bragg} is still the limiting factor.

After the lateral dimensions and thickness is determined, the gap thickness (g) can be chosen. For thermal mechanical current noise limited operation, it is observed that the receiver performance is independent of the DC bias and gap thickness. However, a realistic device integrated with transmit\receive electronics would still require operation close to collapse to increase the signal levels. This limitation makes the maximum available voltage the main limiting factor. For unipolar pulsing, gap thickness is chosen to have close to collapse operation at the maximum available voltage. For bipolar pulsing, DC bias voltage is lower than the maximum available voltage and a smaller gap thickness can be chosen [52]. Smaller membranes require thinner thickness values, enabling larger gap thicknesses for the same operation voltage. As a result, an increase in SNR in addition to the increased bandwidth can be obtained by utilizing smaller membrane geometries.

To illustrate the design process, design of a 1-D CMUT imaging array with a center frequency of $f_{center} = 60$ MHz and a fractional bandwidth of $FBW = 50\%$ is considered. Similar to the analyzed geometries, a fill factor of 64%, and electrode coverage of 56% is chosen. Assuming 60 V operation with unipolar pulsing, the collapse voltage is chosen to be 67 V, to operate at 90% of the collapse voltage. The details of the design process of this array can be seen in Fig. 10 (shown in blue). The resulting parameters shown in the last box in Fig. 10 indicate that the center frequency and fractional bandwidth requirements are satisfied. The pulse-echo response of the designed CMUT array from a perfect reflector at 5 mm distance and 6.7 ns unipolar pulse can be seen in Fig 11.

V. Discussion and limitations

The foregoing design process has also its limitations in terms of practicality. As noted above, while determining the membrane thickness, the membrane f_{ares} needs to be considered to

ensure that f_{Bragg} is the limiting factor. This 2nd symmetric mode (shown as inset in Fig. 12) results in a dip in the far-field transmit pressure. To illustrate the effect of this limitation, in Fig. 12 the SNR (normalized to element area) and BW for the CMUT arrays as a function of membrane width. All CMUT arrays have 40MHz center frequency and 64% fill factor, resulting in f_{ares} and f_{Bragg} crossing over around membrane width of 8.5 μm . When $f_{ares} < f_{Bragg}$, the -6 dB frequency span estimation diverges from $f_{Bragg}/4$. While the bandwidth still increases (beyond 100% FBW shown by the dashed line), the slope is smaller. Our simulations show that for high frequency arrays operating around 30 MHz and above, this condition is encountered when membrane thickness needs to be reduced to below 500 nm. This value is close to the CMUT fabrication limit. Therefore, for practical high frequency CMUT arrays, the proposed design guidelines based on f_{Bragg} are valid and can achieve nearly 100% FBW. Also, it should be noted that the area normalized SNR achieved in the f_{ares} limited region is lower indicating a SNR-bandwidth compromise for high frequency CMUT arrays. These results would be valid for low frequency arrays as well, but practical limitations relevant to f_{ares} can be different for low frequency CMUT arrays where one can have larger lateral size and thicker membranes leading to more than 100% FBW [37], but the SNR compromise in pulse-echo mode should be evaluated.

The reflector distance utilized in the analysis and design process was significantly larger than the array dimensions, resulting in far field operation. In the near field, the effect of the asymmetric vibration modes (Arrow #7 in Figs. 3 and 4) are more prominent which may impact the predicted bandwidth.

Although the discussion here focused on phased array operation with array element pitch close to half wavelength, the design guidelines for bandwidth limits are also valid for larger array elements suitable for linear 1-D arrays. Overall, for larger elements the bandwidth remains the same as it is dominated by the membrane pitch within the element and single membrane dynamics but the beamwidth is reduced as expected. With regards to focused array operation, simulations are conducted for 1mm and 3mm focal distances, F# 3.3 and F#10, respectively. The results indicate that the bandwidth and SNR variations are similar to single element array results. With steering, the bandwidth is reduced proportionally since the linear phase for steering changes the effective Bragg resonance.

The results presented here apply for non-collapsed CMUT operation. This is especially relevant when the lateral dimensions of high frequency phased array elements are considered which need to be close to the half wavelength of operation frequency. Arrays operating in collapse-mode generally require larger membrane dimensions than arrays operating in conventional mode [54]. Consequently, the membrane dimensions operated in collapse-mode in literature are generally above 25 μm , half wavelength at 30 MHz [30]. The constraint on membrane dimensions coupled with the thin membrane requirements may lead to prohibitively large voltages and reliability issues in collapsed mode. However, we note that similar Bragg resonance frequency response limitations should apply to collapsed mode operated CMUT arrays in general, as it is a fundamental phenomenon due to array periodicity. Similarly, CMUT or PMUT arrays with different membrane shapes, such as circular membranes, have the similar limitations due to Bragg resonance and higher

membrane vibration modes (see for example the plots Fig. 6 for higher order mode and $1.0 \lambda_0$ and $0.8 \lambda_0$ pitch 2D arrays in Fig. 7 for Bragg resonance effects in [48]).

VI. Conclusion

A comprehensive receiver and pulse-echo performance analysis for high frequency 1-D CMUT arrays is described and implemented. The pressure noise spectrum is obtained from thermal mechanical current noise and plane wave pressure sensitivity and used to investigate receiver performance of different 1-D high frequency CMUT imaging array geometries with 40 MHz center frequency. It is observed that Bragg's resonance defined the upper limit and crosstalk induced array modes defined the lower limit for most of the design space. The receiver performance between these two limits can be comparable to that of an ideal piston. Reducing the membrane lateral dimensions and pitch improved the minimum detectable pressure and frequency range. Pulse-echo response of the arrays is also analyzed for a planar reflector positioned in far field to obtain SNR values. Similar to receiver characteristics, with reduced membrane dimensions and pitch, higher bandwidth and SNR was obtained. Finally, based on the results obtained from the receiver and pulse-echo analysis a design process is proposed. The array pitch, lateral dimensions and thickness of the membrane are set based on the Bragg's resonance and array crosstalk for the desired frequency response. Gap thickness is then adjusted for these lateral dimensions and available voltage to obtain maximum SNR. Analysis of the design space shows that the design guidelines are valid for high frequency CMUT arrays with practical membrane dimensions.

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APPENDIX

DC BIAS AND GAP THICKNESS

EFFECT ON RECEIVER PERFORMANCE

The receiver performance analysis is extended to analyze the effect of DC bias as percentage of the collapse voltage, and gap thickness. The array geometry populated with $20 \mu\text{m}$ square membranes is considered as the basis. The DC bias is varied from 50% to 90% collapse for an array with 45 nm gap thickness, and the gap thickness is varied from 25 nm to 65 nm while the DC bias is kept at 90% collapse. Comparison of the extracted thermal mechanical current noise, plane wave pressure sensitivity and pressure noise spectrum of a single element of the imaging array can be seen in Fig. A1 and Fig A2.

As expected, both pressure sensitivity and thermal mechanical current noise increase as the DC bias is increased, with the effect of spring softening visible across the frequency spectrum. The same trend is observed as the gap thickness is decreased. Analyzing the pressure noise spectrum of both cases reveals that the increase in pressure sensitivity is mostly nulled by the increase of current noise for both parameters. The effect of spring softening is visible in the pressure noise spectrum, but is limited with respect to the overall receiver performance. From these results, it can be deduced that the gap thickness and to a

lesser extent the DC bias have limited effect on the receiver performance of thermal mechanical current noise limited CMUT array element. As a result, the receiver performance is dictated by the membrane and array geometry, simplifying the design process.

OPERATION VOLTAGE EFFECT ON

PULSE-ECHO RESPONSE

The pulse-echo analysis is extended to analyze the effect of operation voltage. The array geometry populated with 20 μm square membranes is considered as the basis. Unipolar pulsing at 40, 60 and 80 V DC bias is carried out. The gap thickness and HfO_2 isolation thickness are adjusted to obtain 90% collapse and 75% of the breakdown field at each operation voltage. Larger DC biases enable utilization of larger gap thicknesses, resulting in larger pressure output and lower current noise. A comparison of the obtained reflection currents and the thermal mechanical current noise at the different operation voltages can be seen in Fig. B1. The characteristics of the reflection currents and the resulting pulse-echo SNR values can be found in Table B1. The results clearly demonstrate that increasing the operation voltage increases the SNR without affecting the frequency response of the array. From these results, it can be inferred that maximum available voltage must be utilized for best pulse-echo response. The DC bias of the CMUT array should be maximized with respect to the pulsing method and electronics. The gap thickness should then be maximized accordingly. As the receiver performance is relatively independent of the gap thickness, available DC bias can be chosen as the main determinant of this design parameter.

Fig. A1.

DC bias effect as percentage of collapse voltage (V_{col}) on thermal mechanical current noise (I_{noise}), pressure sensitivity (PS) and pressure noise spectrum of 20 μm membrane imaging array elements

Fig. A2.

Gap thickness effect on thermal mechanical current noise (I_{noise}), pressure sensitivity (PS) and pressure noise spectrum of 20 μm membrane imaging array elements

Fig. B1.

Operation voltage effect on the reflection current and thermal mechanical current noise of 20 μm membrane imaging array elements

Table B1.

Characteristic of reflection currents and pulse-echo SNR of 20 μm membrane imaging array elements of different operation voltages

Operation Voltage [V]	Gap [nm]	Iso [nm]	Center Frequency [MHz]	Fractional Bandwidth [%]	SNR [dB]
40	36	135	43.7	23	67.8
60	45	200	43.7	23	70.2
80	54	265	43.9	23	71.8

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Fig. 1.
Block diagram of model developed by Satir *et al* [41]

Fig. 2.
2-D representation of CMUT array. The inset shows a cross-section of a single membrane.

Fig. 3. Thermal mechanical noise spectra of single elements (SE) (dashed line) and imaging array elements (AE) (solid line) of different membrane geometries.

Fig. 4. Pressure noise spectra of ideal pistons (black dashed line), single elements (SE) (dashed colored line) and imaging array elements (AE) (solid line) of different membrane geometries

Fig. 5.
Membrane geometry effect on the reflection current of imaging array elements

Fig. 6.
Membrane geometry effect on the beam pattern of array elements

Fig. 7. Plane wave sensitivity spectra of single elements (SE) and imaging array elements (AE) of different fill factor (FF) array geometries

Fig. 8.
Fill factor (FF) effect on reflection current of 10 μm membrane imaging array elements

Fig. 9. -6 dB frequency span (BW) relation with f_{Bragg_ave} . The inset table shows the characteristics of the additional simulated array geometries and their pulse-echo responses.

Fig. 10. Design algorithm for 1-D high frequency imaging array of $f_{center}=60 \text{ MHz}$ and $FBW=50\%$

Fig. 11.
Reflection current of imaging array element determined from the design process

Fig. 12.

Pulse-echo characteristics of f_{Bragg} and f_{ares} limited arrays. The element pitch varies between 15.6 μm and 25 μm and SNR is normalized to element area.

Table 1.

CMUT membrane geometries of receiver analysis

50×50 μm ² section	a [μm]	d _x [μm]	d _y [μm]	d _{el} [μm]	# of Elems	h [μm]	g [nm]
10	12.5	12.5	25	12	0.6	72
15	18.75	18.75	18.75	16	1.2	60
20	20	25	25	12	2.2	45

Table 2.

Characteristic of reflection currents and pulse-echo SNR of different membrane geometries in water

Membrane Width [μm]	Center Frequency [MHz]	Fractional Bandwidth [%]	Pulse-echo SNR [dB]
10	43.4	69	72.0
15	41.6	41	67.8
20	43.7	23	70.2

Table 3.

Characteristic of reflection currents and pulse-echo SNR of different membrane geometries in blood

Membrane Width [μm]	Center Frequency [MHz]	Fractional Bandwidth [%]	Pulse-echo SNR [dB]
10	33.3	82	57.3
15	36.4	55	53.8
20	40.7	33	55.1

Table 4.

CMUT membrane geometries of fill factor analysis

50×50 μm ² section	a [μm]	d _x [μm]	d _y [μm]	d _{el} [μm]	FF [%]
10	12.5	12.5	25	64
10	11	12.5	25	73
10	10	12.5	25	80

Table 5.

Characteristic of reflection currents and pulse-echo SNR of 10 μm membrane imaging array elements of different fill factor (FF)

Fill Factor [%]	Center Frequency [MHz]	Fractional Bandwidth [%]	Pulse-echo SNR [dB]
64	43.4	69	72.0
73	44.0	74	72.5
80	44.4	78	72.9