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Phase and Amplitude Modulation Methods for Nonlinear Ultrasound Imaging with CMUTs

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Abstract

Conventional amplitude and phase modulated pulse sequences for selective imaging of nonlinear tissue and ultrasound contrast agents are designed for piezoelectric transducers that behave linearly. Inherent nonlinearity of CMUTs, especially during large signal operation, renders these methods inapplicable. In this work we present different pulse sequences for nonlinear imaging that are valid for small- and large-signal CMUT operation. For small-signal operation, two-pulse amplitude and phase modulation methods for microbubble and tissue harmonic imaging are presented, where CMUT nonlinearity is compensated via subharmonic excitation. In the large-signal regime, using a nonlinear model we first show that there is a simple linear relationship between the phases of each harmonic distortion component generated and the input drive signal. Based on this observation, we demonstrate a pulse sequence using $N+1$ consecutive phase modulated transmit events to extract N harmonics of the nonlinear contrast agent echo content uncorrupted by CMUT nonlinearity. Proposed methods assume no a priori information about the transducer, therefore are applicable to any CMUT. The phase modulation method is also valid for piezoelectric transducers and systems with nonlinearities described by Taylor series where same phase relationship between the input signal and the harmonic content is valid. Proof of principle experiments using a commercial contrast agent validate the phase modulated pulse sequences for CMUTs operating in highly nonlinear collapse snapback mode and for piezoelectric transducers.

I Introduction

Nonlinear ultrasonic imaging techniques use the harmonic distortion information extracted from the received echo signals in pulse-echo imaging systems. Harmonic distortion of the ultrasound signal occurs due to the nonlinear mechanical behavior of gas microbubbles in contrast enhanced imaging or nonlinear wave propagation within the tissue in tissue harmonic imaging. Using transmit pulses at the fundamental frequency and the harmonic content within the received echo signals, selective nonlinear images of contrast agents or tissue are constructed. Harmonic imaging offer distinct advantages over linear methods in terms of image resolution, aberration artifacts, clutter, and side and grating lobe levels [1]. Clinical studies also report that patients with features those are not visible in linear ultrasonic images can be examined via harmonic imaging with success [2].

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Successful extraction of nonlinear echo content is crucial for adequate harmonic imaging performance since the echo signals contain both linear and nonlinear reflections, where nonlinear echoes may be may be ~ 30 dB below linear echo levels [1]. A variety of multi-pulse transmit techniques have been developed for separation of nonlinear echoes from linear echo content [3–7]. These methods, such as pulse inversion, rely on inter-pulse phase and amplitude modulation exploiting the linearity of piezoelectric transducers in the expense of image frame rate. By simple summation of multiple echoes, the nonlinear echoes are extracted by elimination of the linear reflection of the transmitted signal at the fundamental frequency. Furthermore in contrast enhanced imaging, it is also essential for an imaging system to be able to distinguish the microbubble echoes from nonlinear tissue reflections. Recently a two-pulse sequence was reported to improve the contrast-to-tissue ratio (CTR) in second harmonic contrast images [8]. The method exploits the Taylor series description of the tissue nonlinearity [9]. Since contrast agent dynamics do not obey such a relation [10], the presented pulse sequence eliminates the tissue signal while maintaining the contrast agent echoes, improving the image CTR. In another study, it has been also reported that CTR improves as a function of the order of the harmonic frequency and by using the higher order harmonic content for image construction 40 dB improvement of CTR over conventional second harmonic imaging has been demonstrated [11].

Unlike piezoelectric transducers, CMUTs exhibit strong nonlinear behavior [12–15]. Nonlinear transducer dynamics render the aforementioned multiple pulse techniques inapplicable to CMUTs. As the transmitted wave is distorted already at the transducer surface, linear echoes and nonlinear echoes cannot be differentiated when received. To overcome this issue, several methods based on pre-distortion of input signals have been proposed to suppress the harmonic content in the transmitted waveforms [14, 16–19]. Utilizing the wide-frequency band operation offered by CMUTs and pre-distorted waveforms, experimental imaging results have been reported, demonstrating the potential of CMUTs in tissue harmonic and contrast imaging [20, 21], however these methods require device calibration and complex drive signals. Earlier, we have shown that the addition of a series impedance suppresses harmonic distortion in CMUT output via gap feedback linearization trading off transmit sensitivity [13]. In another CMUT linearization effort, a checker board array architecture has been introduced to suppress the transmitted second harmonic in the far field at the expense of 3 dB transmit power loss [22]. Novell et al. presented a DC bias modulated three pulse sequence that compensates for CMUT nonlinearity where in vitro images of microbubbles with 30 dB CTR were demonstrated [23]; however the approach is limited to small-signal operation, therefore not valid for large-displacements where the CMUT output pressure is maximized. Recently, an alternative amplitude modulation method was demonstrated for large-signal operation where three alternate firings from two subarrays are utilized to eliminate the CMUT sourced harmonic distortion content in the echo signals [24].

In this work, we analyze nonlinear CMUT behavior in small- and large-displacement regimes and derive pulse sequences for harmonic imaging. The first part of the paper takes a brief look at CMUT nonlinearity for small-displacements using an analytical model. Based on the analysis, we show that pulse inversion can be realized by sub-harmonic excitation with no DC bias and firing two consecutive pulses with $\pi/2$ radians phase difference.

Alternatively, a modified amplitude modulation sequence can be used as suggested by simulation results. In the second part, we use a nonlinear CMUT model to show that the phase shift of the input signal with respect to its envelope shifts the phase of each harmonic distortion product linearly to the order of the harmonic. Based on this observation, an alternative pulse sequence is presented that suppresses any number of particular harmonic content in the received signal sourced by CMUT nonlinearity using proper number of transmit firings with inter-pulse phase modulation while this can be achieved without any CMUT membrane displacement limitations. The method is demonstrated via simulations and validated by initial experiments with a four-pulse sequence for a linear target and microbubbles in an imaging scenario up to the 4th harmonic. The results also suggest that the presented approach is valid in the collapse-snapback mode operation as well, where the CMUT is driven to the physical limits for maximized output pressure [25]. Finally, we present experimental results suggesting the method is also applicable to a system using a piezoelectric transducer, where effect of nonlinearities associated with the transducer, drive electronics and wave propagation may be significantly suppressed in a contrast agent imaging scenario.

II CMUT Operated in Small-Signal Regime: Phase and Amplitude Modulation With Subharmonic Excitation

To develop phase and amplitude modulation schemes in the small signal regime, we perform a simple first order analysis of dynamic CMUT behavior. CMUT transmit operation is described by a linearized lumped spring, mass, piston radiator and parallel-plate capacitor model [12]. The model is linearized for membrane displacements much smaller than the parallel-plate gap and describes the transmitted pressure as a linear function of input voltage squared, i.e. $p(t) = \mathcal{L}\{v^2(t)\}$ where $\mathcal{L}$ is the linear operator that accounts for the linear membrane dynamics and wave propagation within the immersion medium. In a typical harmonic imaging scenario, the input voltage signal consists of a DC term that sets the CMUT sensitivity and a windowed single frequency tone burst,

$$v(t) = V_{DC} + V_{AC} w(t) \cos(\omega_0 t + \varphi). \quad (1)$$

The radiated pressure is then

$$p(t) = \mathcal{L} \left\{ V_{DC}^2 + \frac{V_{AC}^2}{2} w^2(t) + 2V_{DC} V_{AC} w(t) \cos(\omega_0 t + \varphi) + \frac{V_{AC}^2}{2} w^2(t) \cos(2\omega_0 t + 2\varphi) \right\}, \quad (2)$$

Eq. (2) can be reduced to the superposition of 3 terms,

$$p(t) = \frac{V_{AC}^2}{2} \mathcal{L}\{w^2(t)\} + 2V_{DC} V_{AC} \mathcal{L}\{w(t) \cos(\omega_0 t + \varphi)\} + \frac{V_{AC}^2}{2} \mathcal{L}\{w^2(t) \cos(2\omega_0 t + 2\varphi)\}, \quad (3)$$

since no acoustic transmission occurs at DC, i.e. $\mathcal{L}(V_{DC}^2)=0$.

It should be noted that the relationship in (3) is also valid for a gap feedback linearized CMUT where gap dependence of electrostatic force is suppressed by addition a series gap feedback impedance [13]. Therefore by realization of such a linearization scheme, the small-displacement methods presented in this section can be extended to the large-displacements; however the approach is beyond the scope of this work.

Multiple observations can be made by analyzing (3), which are valid for small displacements or a feedback linearized CMUT only. First, only the second harmonic distortion is present in transmission. The second harmonic term can be neglected if $V_{AC} \ll V_{DC}$, suggesting linear device operation [12]. Second is the linear relationship between V_{DC} and the amplitude of the fundamental term for constant V_{AC} . By exploiting this observation, the amplitude of the fundamental component can be adjusted via tuning the DC bias while the second harmonic amplitude is kept at the same level. This observation is exploited in the three-pulse bias modulation method presented in [23]. Another observation is that the signal envelope $w(t)$ generates an additional low frequency signal because of the $v^2(t)$ dependence of the forcing term. Finally, if $V_{DC}=0$ only the second harmonic and the low frequency terms are generated. Therefore, by exciting the CMUT at half the desired operation frequency with no DC bias, desired harmonic operation can be achieved [13], i.e. for

$v(t) = V_{AC} w(t) \cos(\frac{\omega_0}{2}t + \varphi)$, the transmitted pressure becomes

$$p(t) = \frac{V_{AC}^2}{2} \mathcal{L}\{w^2(t)\} + \frac{V_{AC}^2}{2} \mathcal{L}\{w^2(t) \cos(\omega_0 t + 2\varphi)\} \quad (4)$$

In this scenario, phase and amplitude modulation methods can be effectively implemented using two pulses with CMUTs as follows:

Pulse Inversion (PI)

The CMUT is driven with two consecutive pulses at half of the desired frequency $\omega_0/2$ with a phase difference of $\pi/2$ radians. This results in two out-of-phase transmitted signals at the frequency ω_0 such that

$$\begin{aligned} p_1(t) &= \frac{V_{AC}^2}{2} \mathcal{L}\{w^2(t)\} + \frac{V_{AC}^2}{2} \mathcal{L}\{w^2(t) \cos(\omega_0 t)\}, \\ p_2(t) &= \frac{V_{AC}^2}{2} \mathcal{L}\{w^2(t)\} + \frac{V_{AC}^2}{2} \mathcal{L}\{w^2(t) \cos(\omega_0 t + \pi)\}. \end{aligned} \quad (5)$$

For a linear reflector, addition of the two echo signals cancel out the ω_0 term, only leaving the low frequency term; while such cancellation does not hold in the presence of microbubbles or tissue nonlinearities.

Fig. 1(top) presents the simulation results for pulse inversion. The CMUT with properties described in the Appendix is driven with a 2 μ s Hann windowed 2.5 MHz sine wave with 8

V peak value and 0 and $\pi/2$ phase shifts and no DC bias. The graph shows the transmitted pressure spectra for both waveforms as well their sum. It should be noted that harmonic distortion still exists in both transmissions, however the total harmonic distortion is less than -40 dB, indicating reasonable linearity; therefore the simulated scenario is suitable for harmonic imaging. In the sum signal, the fundamental component contribution from CMUT is cancelled out completely while the low frequency pulse remains. Moreover, the third harmonic is also cancelled out, which suggest a similar high order relationship exists between the drive signal and the radiated pressure. This observation is investigated for large signal operation in Section III.

Amplitude Modulation (AM)

Two amplitude scaled, in-phase consecutive pulses with drive signal amplitudes V_{AC} and AV_{AC} result in two transmitted pressure waveforms

$$\begin{aligned} p_1(t) &= \frac{V_{AC}^2}{2} \mathcal{L}\{w^2(t)\} + \frac{V_{AC}^2}{2} \mathcal{L}\{w^2(t) \cos(\omega_0 t)\}, \\ p_2(t) &= \frac{A^2 V_{AC}^2}{2} \mathcal{L}\{w^2(t)\} + \frac{A^2 V_{AC}^2}{2} \mathcal{L}\{w^2(t) \cos(\omega_0 t)\}, \end{aligned} \quad (6)$$

For a linear reflector, subtraction of the scaled signals cancel each other, i.e.

$$p_2(t) - A^2 p_1(t) = 0.$$

Fig. 1(bottom) presents the frequency domain results for this amplitude modulation method for two pulses with amplitudes of 6 V and 8 V, where the fundamental contribution from CMUT is suppressed by 50dB. It should be noted that compared to the pulse inversion case, amplitude modulation did not achieve complete cancellation of the fundamental component nor the low frequency signal. This is because device nonlinearities associated with gap dependence of the electrostatic force are still present even though the CMUT is operated in small-displacement regime.

III Large Signal CMUT Behavior: Phase Analysis

To achieve transmit pressure levels comparable to piezoelectric transducers, CMUT arrays need to be operated close to their physical limits [15]. This requires large signal operation where the electrostatic nonlinearities and the gap dependence of the electrostatic force cannot be neglected, rendering the small-signal methods given in Section II of limited use. To utilize CMUTs for nonlinear imaging effectively, the small membrane displacement limitation should be removed. For this purpose, in this section we investigate the phase relationships between the harmonic distortion products and the input signal that can be exploited for contrast agent imaging with CMUTs using the large signal CMUT array model we have developed earlier [26].

The CMUT can be considered to exhibit weak nonlinearity since the harmonic distortion is a smooth function of input signal amplitude and gradually disappears as the input signal gets smaller and smaller [27]. Therefore the system dynamics can be represented as a Volterra series relationship between input signal and output pressure. Such a description is beyond

the scope of this paper, however a low frequency approximation can give an insight to the harmonic distortion in these transducers, at least in terms of the phase relations with respect to the input signal.

For a CMUT element composed of multiple membranes, inertial and acoustic crosstalk effects can be neglected in low frequencies (frequencies smaller than resonance), resulting in a “memoryless” system, which can be described by Taylor series:

$$p(t) = \sum_{n=0}^M \mathcal{L}\{a_n v^n(t)\}. \quad (7)$$

When the phase of the input signal with respect to its envelope shifted by θ , the phase of each harmonic shifts by

$$\theta_n = n\theta \quad (8)$$

as can be shown via (7) using trigonometric identities [28].

The simple phase relation in (8) also holds when the low frequency limitation is removed. To demonstrate this the same CMUT described in the Appendix is driven with V , $V_{AC}=19$ V, $f_0=5$ MHz and $2 \mu\text{s}$ Hann signal envelope. In this scenario the CMUT is driven close to dynamic collapse, i.e. $V_{DC}=19$ V, $V_{AC}=20$ V results in membrane pull-in. The amplitudes normalized to the fundamental component and phase shifts of the first six harmonics are presented in Fig. 2. The results show that the harmonic distortion is invariant to the phase shift of the input θ with respect to its envelope, while each harmonic content shifts by $n\theta$ as discussed.

IV Phase Modulation Method for Contrast Agent Imaging with Nonlinear CMUTs

The simple relationship between the input signal phase with respect to its envelope and the phases of individual harmonic distortion content can be exploited to cancel any undesired harmonic or the fundamental signal in the received echo waveforms from a linear reflector using input signals applied during two consecutive firings with proper phase shift and then adding the received signals. For example, to cancel the fundamental (first harmonic) in the reflections, one applies the second signal with a relative phase shift $\theta=\pi$, which corresponds to pulse-inversion, or to cancel the second harmonic, a phase shift of $\theta=\pi/2$ is applied. The method is also applicable to cancel out the nonlinear echoes from the tissue in contrast enhanced imaging as explored in [8].

Through equation (8), it can also be shown that by firing N consecutive pulses where the

respective phase of i^{th} firing is $\theta_i = \frac{2(i-1)\pi}{N}$ and adding the resulting pressure waveforms, contributions from the CMUT up to the first $(N-1)^{\text{th}}$ harmonic as well as all harmonics

except the M th harmonic and harmonics of the M th harmonic can be eliminated, enabling contrast agent imaging using multiple harmonics. A similar idea is employed in cancellation of harmonics in polyphase multipath circuits, which also exploits (8) [28].

In the presence of nonlinear contrast agent reflectors, the only components in the sum signal would be the contributions from the nonlinear contrast agent vibrations since the contrast agent response will not cancel out in the sum signal [10]. Moreover, as the tissue obeys the Taylor series relation, the contribution resulting from the nonlinear propagation in the tissue would be also eliminated, potentially improving the contrast agent-to-tissue ratio (CTR) in the constructed nonlinear image [8, 29]. In a super-harmonic imaging scenario where all harmonic content is used in contrast agent imaging for improved CTR [11], the left-over harmonic content (M th, $2M$ th, $3M$ th...) cannot be used in image construction since the CMUT contribution is not eliminated and should be filtered out in post-processing of the received echo signals.

A Simulation Results

The proposed phase modulation method is simulated in large displacement regime for performance evaluation for a three-pulse sequence. In this case the elimination of the fundamental and second harmonic components contributed by the CMUT is desired. The same CMUT of the Appendix is driven with $V_{AC}=19\text{ V}$ and $V_{DC}=19\text{ V}$ at 5 MHz. As mentioned above, this drive level corresponds to conventional non-collapse CMUT operation in the large signal regime with nearly full gap swing. The amplitude spectra for three transmit events with respective phases $\theta_1=0$, $\theta_2=2\pi/3$ and $\theta_3=4\pi/3$ and the sum signal $[p_1(t) + p_2(t) + p_3(t)]/3$ are given in Fig. 3. The simulation results show that the three-pulse phase modulation sequence completely eliminates the contribution of the CMUT in the returned echo for all harmonics except the 3rd harmonic and harmonics of the 3rd harmonic. Moreover, this is achieved without compromising the CMUT pressure output.

B Experimental Results

Experiments were performed to evaluate the proposed phase modulation method using a commercial contrast agent, TargestarTM-P (Targeson Inc., San Diego, CA, USA). A parylene coated 8 membrane single CMUT element was immersed in saline, and a hydrophone (HGL-0400, Onda Corp., Sunnyvale, CA) was used to measure the transmitted pressure 6mm away from the transducer. The frequency response of the test device is shown in Fig. 4. Fundamental frequency for the experiments was selected as 5 MHz, so the first four harmonics would be within the 6 dB operation band. A four pulse sequence was implemented, $N=4$, where four consecutive pulses with respective phases $\theta_1=0$, $\theta_2=\pi/2$, $\theta_3=\pi$ and $\theta_4=3\pi/2$ were transmitted and recorded with the hydrophone. This number of pulses is similar to other pulse sequences used in contrast enhanced imaging [5].

The experimental evaluation of the method for non-collapse CMUT operation in the large signal regime was not possible due to the small size of the CMUT element used, which resulted in poor SNR in the recorded waveforms when contrast agent was injected into the immersion saline. Therefore in order to enhance the contrast agent response in the experiments, the transmitted pressure was maximized by operating the CMUT in the

collapse-snapback operation where DC bias is set close to collapse voltage and a large AC excitation signal was applied [25].

First, the pulse sequence is applied without the contrast agent present. The amplitude spectra of the recorded transmitted pressure waveforms for four consecutive pulses and their sum are presented in Fig. 5(top). The results show that the four-pulse sequence suppressed the first, second and the third harmonics by 50dB, 60 dB, and 40dB respectively. In this case, in addition to the CMUT nonlinearity, nonlinearities associated with the RF power amplifier driving the transducer (Model350L, E&I Ltd., Rochester, NY) and wave propagation in saline are present, which are also eliminated in the sum signal. The results demonstrate that the method is valid even for collapse-snapback operation.

To see the effect of contrast agent in the transmitted pressure, the contrast agent is injected into the immersion saline with dilution factor of 1/5000 and the pulse sequence is repeated. The spectra of the recorded waveforms and their sum are shown in Fig. 5(bottom). The overall transmitted signal is attenuated due to the strong backscatter from the contrast agent as expected, but much stronger signals at the fundamental and up to the 3rd harmonic are present in the sum signal as compared to the no-contrast agent case. The results show that the nonlinear content originating from the contrast agent can be extracted with about 20 dB SNR (or similarly CTR assuming tissue harmonic suppression as well as CMUT harmonics) for the first three harmonics using the four-pulse sequence.

To show that the phase modulation method can be used on piezoelectric transducers, similar to Pasovic et al. [8] but extending it to arbitrary number of harmonics, the experiment is repeated using a piezoelectric transducer (7.5 MHz immersion transducer Model IS0702GP, Valpey Fischer Corp., Hopkinton, MA). The transducer is first immersed in saline and excited with a 6 MHz four-pulse sequence with $\pi/2$ radians inter-pulse phase shift using an RF power amplifier driving the transducer. The recorded amplitude spectra for four transmit events and their sum are shown in Fig. 6(top). For each transmission, the second harmonic is ~30 dB and the third harmonic is ~40 dB below the fundamental level. As the echoes from contrast agent may be ~30 dB below the linear echo levels as discussed in Section I, a contrast agent imaging scenario would result in a poor CTR with this configuration. The sum signal, on the other hand, shows that the first three harmonics are suppressed to the noise level, nearly complete elimination of the nonlinear content which may be due to wave propagation, drive electronics as well as the transducer. The experiment is repeated after injection of the TargestarTM-P contrast agent with 1/5000 dilution. The recorded spectra for each transmission and their sum are given in Fig. 6(bottom). In this case the summation does not eliminate the signal, and by comparison with no-contrast agent case ~30 dB SNR at the fundamental frequency and ~20 dB SNR at the second harmonic are achieved.

Conclusion

Several conclusions can be drawn from the work presented here. When the CMUT displacements are small so that the dependence of the output pressure to instantaneous gap can be ignored, there are several strategies to use CMUTs for tissue harmonic and contrast harmonic imaging. As an alternative to the three pulse sequences in literature, two pulse

methods with no DC bias can be used; however the performance of these methods critically depends on the small displacement assumption. As the large signal CMUT modeling reveals, larger signals result in complex interactions such as generation of strong third harmonics which can change the fundamental signal levels in return. Therefore small signal methods need to be carefully evaluated depending on the system output pressure requirements, and the choice of method can depend on system capabilities and ease of implementation. We note that the small displacement limitation for these two pulse methods can be removed using gap feedback linearization of CMUTs, which will be explored further in future work. The fact that CMUT behavior obeys the Volterra series relationship between input and output, such as tissue harmonic generation, is a significant realization, because there is no analytical expression for CMUT behavior to verify this proposition and certain nonlinear systems, such as contrast agent dynamics do not obey such a relation. Once the Volterra series relation is established, it leads to multi-pulse phase modulation methods to reduce the effects of CMUT harmonics without any displacement limitations. The approach can reduce the effect of nonlinearities of the system as well, while contrast agent signals are preserved over a broad bandwidth while tissue signals are suppressed. This is achieved without compromising output pressure levels as the experiments indicate satisfactory harmonic cancellation even for CMUTs operated in collapse-snapback mode. We note that although these large signal methods seem promising, linear or linearized CMUTs still seems to be a requirement for tissue harmonic imaging as tissue signals are lost when phase shifted signals are added. Future work will include comparison of the proposed multi-pulse methods and exploring their limitations through realistic contrast and tissue harmonic imaging experiments.

Appendix: CMUT Model Used in Simulations

For simulation of the proposed methods in this paper, we used the MATLAB/Simulink (The MathWorks Inc., Natick, MA) based nonlinear CMUT array model presented in [26]. As a generic CMUT device, a single CMUT element consisting of 16 square membranes in 4×4 array formation is modeled with parameters given in Table I. The collapse voltage of the modeled CMUT is 32 V. The simulated spectrum of the radiated pressure 1 mm away from the element for 24 V DC bias and 14 V, 60 ns Hann pulse is shown in Fig. A1. For harmonic analysis, the fundamental frequency is chosen as 5 MHz, so the spectrum up to the 4th harmonic is within the 6 dB frequency band of the CMUT.

Fig. A1.
Frequency response of the simulated CMUT

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Fig. 1.

(Top) PI: Pressure output amplitudes for two pulses with $\pi/2$ phase difference and their sum in frequency domain (Bottom) AM: Amplitude spectra of pressure outputs for two pulses with scaled amplitudes and their scaled sum in frequency domain.

Fig. 2.

(Top) Amplitudes of the first six harmonics normalized to the fundamental component as a function of phase shift in input (Bottom) Phase shift of the first six harmonics as a function of phase shift in input.

Fig. 3.

Simulated amplitude spectra of three phase modulated pressure outputs for large signal operation $\theta_1 = 0$, $\theta_2 = 2\pi/3$ and $\theta_3 = 4\pi/3$, and the average signal $|p_1(t) + p_2(t) + p_3(t)|/3$

Fig. 4.
Frequency response of the CMUT used in experiments

Fig. 5.

Amplitude spectra of four phase modulated pressure outputs, $\theta_1 = 0$, $\theta_2 = \pi/2$, $\theta_3 = \pi$ and $\theta_4 = 3\pi/2$ and their average when the CMUT is immersed in saline (top) and saline with microbubbles (bottom)

Fig. 6.

Amplitude spectra of four phase modulated pressure outputs, $\varphi_1 = 0$, $\varphi_2 = \pi/2$, $\varphi_3 = \pi$ and $\varphi_4 = 3\pi/2$ and their average when the piezoelectric transducer is immersed in saline (top) and saline with contrast agent (bottom)

TABLE I

Simulated CMUT Parameters

Membrane Dimensions	$35\mu\text{m} \times 35\mu\text{m} \times 1.65\mu\text{m}$
Electrode Coverage	$25\mu\text{m} \times 25\mu\text{m}$
Gap	100 nm
Array pitch	45 μm
Electrode isolation thickness	100 nm
Isolation dielectric constant	15
Membrane Density, Poisson's ratio and Young's modulus	2040 kg/m ³ , 0.22, 110 GPa