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A 256×256 2-D array transducer with Row-column addressing for 3-D Rectilinear Imaging

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Abstract

We present simulation and experimental results from a 5 MHz, 256×256 2-D (65,536 elements, $38.4 \text{ mm} \times 38.4 \text{ mm}$) 2-D array transducer with row-column addressing. The main benefits of this design are a reduced number of interconnects, a modified transmit/receive switching scheme with a simple diode circuit, and an ability to perform volumetric imaging of targets near the transducer with transmit beamforming in azimuth and receive beamforming in elevation. The final dimensions of a transducer were $38.4 \text{ mm} \times 38.4 \text{ mm} \times 300 \text{ }\mu\text{m}$. After prototyping a row-column transducer, the series resonance impedance was $104 \text{ }\Omega$ at 5.4 MHz. The measured -6 dB fractional bandwidth was 53% with a center frequency of 5.3 MHz. The SNR at the transmit focus was measured to be 30 dB. At 5 MHz, the average nearest neighbor crosstalk was -25 dB. In this paper, we present 3-D images of 5 pairs of nylon wires embedded in a clear gelatin phantom and of an 8 mm diameter cylindrical anechoic cyst phantom acquired from a 256×256 2-D array transducer made from a 1–3 composite. We display the azimuth and elevation B-scans as well as the C-scan. The cross-section of the wires is visible in the azimuth B-scan while the long axes can be seen in the elevation B-scan and C-scans. The pair of wires with 1 mm axial separation is discernible in the elevational B-scan while all the pairs of wires were distinguishable in the short-axis B-scan. Using a single wire from the wire target phantom, the measured lateral beamwidth was 0.68 mm and 0.70 mm at 30 mm depth in transmit beamforming and receive beamforming respectively compared to the simulated beamwidth of 0.55 mm. The cross-section of the cyst is visible in the azimuth B-scan while the long axes can be seen in the elevation B-scan and C-scans as a rectangle.

I. Introduction

A recent major innovation in ultrasound imaging technology is 3-D imaging. 3-D ultrasound has several advantages over 2-D ultrasound, providing orientations and slices not available with 2-D ultrasound, more accurate volume measurements and improved detection of cystic or cancerous masses [1]. For example, 2-D breast ultrasound is routinely used for identifying cysts and solid nodules [2]. 3-D ultrasound has the potential to demonstrate the delineation of these lesions and to reduce the need for biopsy [3]. When a biopsy becomes necessary, 3-D ultrasound-guided breast biopsy procedures may help doctors by giving 3-D spatial location of the applied needle since the needle no longer has to be aligned within a single scan plane [4]. Another application of 3-D ultrasound is ultrafast ultrasound in vascular imaging [5]. This 3-D vascular system can be expected to improve assessment of luminal geometry and disease process, since a 3-D data set can be freely rotated and examined along multiple planes and tomographic sections.

Current 3-D ultrasound systems may contain mechanically moving 1-D array transducers or fully or partially populated 2-D array transducers. Volumetric images are then captured by physical movement of 1-D array or by electronic scanning in the case of a 2-D array. Mechanical translation of a 1-D array has a slow data acquisition rate, poor spatial resolution in the elevation direction, which in turns degrades image quality and limits accuracy of the volume measurement [1]. A fully-sampled 2-D array consisting of $128 \times 128 = 16,384$ or 256

$\times 256 = 65,536$ elements would serve as an ideal probe for 3-D rectilinear ultrasound imaging of near-field targets such as breast, carotid artery and abdomen. Such a fully sampled 2-D array would give a substantial improvement over a traditional 1-D linear array in image quality in 3-D imaging with a wide field of view near the transducer. However, it has not been demonstrated whether developing a fully sampled 2-D array would be feasible because of the difficulty in fabricating an array with such a large number of elements. The high electrical impedance due to the small size of each element would result in signal loss. To handle such a large array, one would need as many as 65,536 channels and tens of thousands of coaxial cables in an imaging system create a major challenge.

One well established technique to overcome these difficulties is by using a sparse 2-D array [6–8]. Yen and Smith investigated several sparse 2-D array designs for real time volumetric imaging [6] and used the Mills-cross design to build a curvilinear array for 3-D ultrasound [7]. In the later version of the rectilinear 3D scanner, a sparse periodic array with receive mode multiplexing was built to improve spatial resolution performance [8]. However, extreme sparseness, with less than 10 % active elements, causes grating lobes and high clutter levels. This results in degradation of image contrast and hence limits the diagnostic value of the exam. Taking advantage of integrated circuit (IC) fabrication techniques, recent commercial systems use integrated electronics to allow connecting to approximately 3,000 individual elements before funneling the signals from these elements into 128 system channels via switches and preliminary beamformation within the handle [9]. The preliminary beamformation is located within the transducer handle using application specific integrated circuits (ASICs). The second stage of beamformation is done by a traditional 128 channel system digital beamformer providing dynamic focusing and full aperture delays. Tamano et al. developed a convex 2-D array in which switches connect elements in rings around the beam center similar to an annular array [10,11]. Changing a steering direction or transmit focus is accomplished by reconfiguring the switches.

Another emerging array technology is capacitive micromachined ultrasonic transducers (CMUTs). This technology is suitable for 3-D imaging with 2-D arrays since the challenges associated with small element size in conventional array fabrication are not present. CMUT technology makes use of advanced IC fabrication processes and enables the easy manufacture of large 2-D transducer arrays with individual electrical connections. With through-wafer interconnects and a 420 μm element pitch, Oralkan et al. fabricated and demonstrated this technology with an experimental prototype, a 32×64 -element portion of the 128×128 -element array [12].

Ermert et. al. presented an ultrasound transmission camera using two separate linear crossed arrays [13]. An array consisting of long horizontally oriented transducers, which are focused vertically is used in transmit. In receive, the orientation of the second array is vertical which can be focused horizontally. This transmission ultrasound system has real-time capability and a reduced number of channels. It is an alternative to an earlier design of transmission camera with a 2-D array. Morton and Lockwood presented a crossed-electrode array that has two identical hemispherically shaped linear array electrode patterns oriented perpendicular to each other on opposite sides of a 1–3 composite [14]. Using this design, transmit beamforming in one direction and receive beamforming in the other direction could be achieved. This transducer acquires a pyramidal volume by emitting a fan shaped beam suitable for cardiac imaging. In similar work, we presented a row-column addressing technique to simplify interconnections of a flat 40 mm $\times$ 40 mm 2-D transducer array and verified its performance through simulations and experiments [15]. This transducer array is essentially a 1–3 composite with vertical and horizontal electrodes on the top and bottom respectively. Transmit and receive switching between the vertical and horizontal electrodes was accomplished with a simple diode circuit. In another realization of the row-column or cross-electrode array design, we proposed a dual-

layer transducer array design which uses perpendicular 1-D arrays for 3-D imaging [16]. This dual-layer design uses one piezoelectric layer for transmit and another separate co-polymer layer for receive. Each layer is an elongated 1-D array with transmit and receive elements oriented perpendicular to each other. Both of these row-column transducers scan the volume rectilinearly with a wide field of view close to the transducer and are useful for abdominal, breast, or vascular imaging.

In this paper, based on our previous works with Mills cross and row-column addressing scheme, we present simulation and experimental results from a 5 MHz 256×256 (38.4 mm $\times$ 38.4 mm) 2-D array transducer for rectilinear volumetric imaging. This paper is organized as follows. Section II presents the design of the row-column addressing scheme. A point target simulation is performed to evaluate the lateral resolution and clutter level. The fabrication and testing methods as well as experimental setup are described in section III. Experimental results are presented in section IV, and section V discusses the results and concludes the paper.

II. DESIGNS

A. Transducer design

First, we describe this design using a simplified 8×8 2-D array (Fig. 1). This design utilizes a two-layer electrode pattern where the bottom layer consists of a series of vertical electrodes (Fig. 1A) and the top layer consists of a series of horizontal electrodes (Fig. 1B). In transmit, the vertical electrodes serve as the “ground” and the top electrodes serve as the “transmitters”.

In this example, we excite transmit channel D as indicated by the arrow to the right of channel D in Fig. 1B. This row of elements, shown in the gray shading in Fig. 1C, then emits a cylindrical wavefront into the field. In elevation, the wavefront appears omnidirectional since the aperture behaves like a single small element. In the azimuth direction, the emitted beam is a planar wavefront because all elements fire simultaneously and the aperture behaves as a single long element. For receive mode, receive channels A-H are active and the desired receive column is selected (Fig. 1D). In receive mode, the individual elements along one column (gray shading) will be used to record the echoes (Fig. 1F). With this design, transmit beamforming can be done in the vertical or elevational direction while receive beamforming can be done horizontally or azimuthally. A schematic illustrating this beamforming method is shown in Fig 2. Multiple rows could be used for elevational beamforming in transmit and multiple columns can be used for azimuth beamforming in receive. By stepping transmit subapertures across the array with multiple receive beams within the transmit beam, a 3-D rectilinear image can be achieved [14].

To implement this design, we have modified a typical transmit/receive circuit such that an entire row acts as a single transmit element and an entire column acts as a single receiver [17,18]. A typical Tx/Rx circuit uses the same electrode for ground in both transmit and receive. The modified Tx/Rx circuit uses one electrode as ground in transmit and the other electrode as ground in receive (Fig. 3b) Since the effective size of the element is increased by a factor of N (where N is the number of elements in one direction in the $N \times N$ 2-D array transducer), the high electrical impedance associated with a single 2-D array element is eliminated. This modified transmit/receive circuit is illustrated in Fig. 3a) using a simpler 2×2 array.

In this circuit, the row electrodes act as transmitters and are connected to the transmitters. The column electrodes are not connected directly to ground, but a pair of parallel reversed diodes serves as a low impedance path to ground. When either Tx1 or Tx2 emits a high voltage transmit pulse along the row electrodes, approximately ± 50 – 100 V, one of the diodes in each pair is forward biased with a ± 0.7 V drop with respect to ground. The remaining voltage drops across the transducer, and the column electrodes serve as a ground through the forward-biased diodes.

In receive, the relatively low output impedance of the transmitter and a 1 k Ω pull-down resistor (R1) provide a path to ground and the diodes are turned off assuming none of the echoes have an magnitude greater than 0.7 V. Thus, the row electrodes effectively become the ground electrodes. Using this row-column addressing scheme, we performed a point spread function simulation using Field II [19] (Fig. 4). The transmit aperture is a 1-D array with azimuthal pitch of $\lambda/2 = 0.15$ mm and elevational height of $256 \lambda/2 = 38.4$ mm. The receive aperture has an elevational pitch of $\lambda/2 = 0.15$ mm and an azimuthal length of 38.4 mm. A Gaussian pulse with center frequency of 5 MHz and a -6 dB fractional bandwidth of 50 % was used. For cross-sectional scanning, a 128-element subaperture was used in both transmit and receive and focused on-axis to (x,y,z) = (0,0,30) mm. The highest clutter levels, around -30 to -40 dB, are seen along the azimuth and elevation axes. The -6 dB and -20 dB beamwidths are 0.55 mm and 2.39 mm respectively.

III. Methods

A. Array fabrication

The 1-3 composite device was manufactured using the conventional dice and fill technique [20]. To fabricate this 2-D array substrate, a 50 mm $\times$ 50 mm wafer of 600 μ m thick PZT-5H was mounted on a 3 $\times$ 3 inch glass plate using melted wax. The PZT was diced in perpendicular directions to a depth of 325 μ m with a 150.23 μ m element-to-element distance and a 35 μ m kerf. The kerfs were filled with Epotek 301, a low viscosity epoxy. This initial pitch size of 150.23 μ m was calculated to take into account epoxy shrinkage during the fabrication process to achieve a final pitch of 150 μ m. Next, any excess epoxy was lapped away and gold and chrome layer was sputtered on the top side. A diamond wheel dicing saw was used to scratch dice a few microns into the epoxy filler to form columns. The PZT was then flipped over and lapped to a final desired thickness near 300 μ m. After sputtering the bottom side, row electrodes were created by scratch dicing. The final array pitch is equal to the composite pitch, or 150 μ m, where a line of square ceramic pillars is connected by a single electrode, forming one array element. Two prototype flexible circuits (Microconnex, Snoqualmie, WA), oriented perpendicular to each other were bonded on opposite faces of the composite (Fig. 5) using a pressure of approximately 100 psi. The flexible circuits consist of a 104 μ m thick polyimide, 9 μ m thick copper, 1 μ m thick nickel and 1 μ m thick gold. The polyimide layer of the flexible circuit has an acoustic impedance of 3.4 MRayl and was attached to the front of the transducer as a matching layer. Polymer films have been previously used as a coupling matching layer previously [21]. The flexible circuit was connected to a custom printed circuit board (PCB) (Sunstone Circuits, Mulino, OR) by a Samtec connector (Samtec, Inc. New Albany, IN). The parallel reversed diode pairs (Fig. 3) are located on the receive PCB.

A 12 mm thick lossy epoxy backing consisting of 45 g of tungsten oxide (Cerac Specialty Inorganics, Milwaukee, WI), 7.875 g of LP-3 (Morton International, Chicago, IL), 1.688 g of phenolic balloons (Union Carbide, Danbury, CT), 4.5 g of Zeeospheres (3M, St. Paul, MN), 14.625 g of Dow Epoxy Resin 332 and 5.063 g of Dow Epoxy Hardener 24 (Dow Chemical Company, Midland, MI) was bonded onto the flexible circuit behind the transducer. The calculated acoustic impedance was 4.5 MRayl and the attenuation in the backing was measured using the method in [22] and was 10 dB/cm at 5 MHz.

B. Measurements

We have verified the operation of the modified circuit (Fig 3A) experimentally using the Ultrasonix Sonix RP ultrasound system (Ultrasonix Medical Corporation, Richmond, Canada,) with a Panametrics 5 MHz piston transducer (Fig 3B). In a standard setup, the echo was received from a glass plate using a standard Tx/Rx channel from the Ultrasonix RP system, a 5 MHz Panametrics piston transducer and a pair of parallel, reversed diodes. In a switched case, one

channel was used as a transmitter only and another channel is used in receive only to receive echo. The voltage trace is measured across a pair of parallel, reversed diodes on the receive board.

An Agilent (Santa Clara, CA) precision impedance analyzer (model 4294A) was used for impedance analysis. Since the individual pillars of the composite are not individually accessible, all electrodes of one side from the transducer were grounded while the other side was used to measure electrical impedance. The device is similar to an elongated 1-D array. Pulse-echo measurements, using an aluminum plate reflector and a Panametrics 5072PR pulser/receiver (Waltham, MA) were taken in the same manner as for the impedance measurements. The pulse was acquired using a Tektronix oscilloscope (TDS 5054) and Fast Fourier Transform done in Matlab (Mathworks, Natick, MA) to yield the spectrum. Nearest-neighbor crosstalk measurements of the arrays were made by applying a 200 mV_{p-p} 5 MHz 20-cycle burst using an Agilent 33250A function generator on one element and measuring the voltage on the neighboring element with 1 M Ω coupling on the oscilloscope. The system signal-to-noise ratio (SNR) at the transmit focus was measured using the method described in [23]. The composite transducer array was interfaced with the Sonix RP ultrasound system using a custom printed circuit board to acquire data from a tissue mimicking phantom. Next, the transmitters were turned off to acquire only electronic noise data. After low-pass filtering, standard beamforming, envelope detection and log compression, the difference between a signal mean and a noise mean image was calculated to get the system SNR.

C. Imaging Experiments

After transducer testing, the composite transducer array was interfaced with the Sonix RP ultrasound system using a custom printed circuit board. This system allows the researcher to control imaging parameters such as transmit aperture size, transmit frequency, receive aperture, filtering, and time-gain compensation. In these experiments, one row was connected to one channel of the Sonix system. This channel was used in transmit mode only. A two-cycle, 5 MHz transmit pulse was used. Sixty-four receive columns were each connected to individual system channels configured to operate in receive mode only. With a 40 MHz sampling frequency, data from each receive channel was collected 100 times and averaged to minimize effects of random noise. A different set of 64 receive elements was used until data from all 256 receive elements were collected. This process is repeated until all transmit and receive element combinations were acquired. The data was then imported into Matlab for offline 3-D delay-and-sum beamforming, signal processing and image display. Dynamic transmit (azimuth) and receive (elevation) focusing was done with 0.5 mm increments with a constant f-number of 2. The image line spacing was 150 μ m. All signals in the experiments were bandpass filtered using a 64-tap finite impulse response (FIR) bandpass filter with frequency range of 3.75 – 6.25 MHz. A 3-D volume was acquired by moving the transmit subapertures in azimuth and receive subapertures in elevation to focus a beam directly ahead.

We imaged homemade 70 $\times$ 70 $\times$ 70 mm tissue mimicking gelatin phantoms containing 5 pairs of nylon wire targets with axial separations of 0.5, 1, 2, 3 and 4 mm. The bottom wire in each pair was shifted laterally by 1 mm with respect to the top wire. The diameter of the nylon wire was 400 μ m. The background material of the wire phantom consisted of 400 g of DI water, 36.79 g of n-propanol, 0.238 g of formaldehyde and 24.02 g of gelatin. These ingredients and quantities are based on recipes given in the literature for evaluating strain imaging techniques [24]. The second phantom imaged was an 8 mm diameter cylindrical anechoic cyst phantom where the cyst was located at a depth of 30 mm. The cylindrical cyst was made using the same ingredients as the background of the wire phantom. The tissue mimicking material surrounding the cyst used the same ingredients as the wire phantom but with 3.89 g of graphite powder added to provide scattering.

The dimensions of the acquired volumes were 40 (azimuth) $\times$ 40 (elevation) $\times$ 45 (axial) mm. After beamforming, envelope detection was done using the Hilbert transform. Images were then log-compressed and displayed with a dynamic range of 20 to 30 dB. Azimuth and elevation B-scans are displayed along with C-scans which are parallel to the transducer face.

IV. Experimental Results

Fig. 6 shows the photograph of the final transducer. The final pitch was 150 μ m. Two perpendicular flexible circuits can also be seen in this figure.

Fig. 7 shows the pulse-echo result from the experiment to verify the operation of the modified circuit (Fig 3). The solid line in Fig. 7 is the echo received from a glass plate using a standard Tx/Rx channel from the Ultrasonix RP system, a 5 MHz Panametrics piston transducer and a pair of parallel, reversed diodes. The dashed line is the echo where one channel was used as a transmitter only and another channel is used in receive only. Since the ground of the transducer is switched between transmit and receive, the echo using the new circuit is inverted compared to when a traditional transmit/receive circuit is used.

Fig. 8 shows the typical electrical impedance of a transducer element measured experimentally using an Agilent precision impedance analyzer and in simulation using the 1-D KLM model [25]. The simulated impedance magnitude was 100 Ω at a series resonance frequency of 5.3 MHz while the experimental impedance curve showed a series resonance of 104 Ω at 5.4 MHz. The phase plots peak at 6.4 MHz for the KLM simulation and at 6.1 MHz in the experimental case.

Fig. 9 shows the simulated and experimental time and frequency responses of the pulse-echo signals. In simulation, the center frequency was 5.2 MHz with a -6 dB fractional bandwidth of 67 %. Experimentally, the spectrum of the pulse has a center frequency of 5.3 MHz and a -6 dB fractional bandwidth of 53 %.

At 5 MHz, the average nearest neighbor crosstalk was -25 dB. This level of crosstalk is acceptable in use for conventional linear array imaging, where the acoustic beam is not steered [26]. The SNR at the transmit focus at a depth of 30 mm (f -number = 1.5) was measured to be 30 dB [23].

Fig. 10A-C show the azimuth B-scan, elevation B-scan, and C-scan of the wire phantom respectively when the short axis of the wires is in the azimuth direction. All images are log-compressed and shown on a 20 dB dynamic range. The elevation B-scan (Fig. 10B) shows the pair of wires with 1 mm axial separation. The two wires are discernible. The C-scan was taken at a depth of 35 mm. Here, one can also see the presence of sidelobes along side the wires. Fig. 10D-F show the axial wire target phantom with the short axis of the wires in the elevation direction. The pair of wires with 1 mm axial separation is discernible in the azimuth B-scan while the short-axis view is shown in Fig. 10E. Fig. 10F shows the C-scan where sidelobes are again present.

Fig. 11 shows the lateral wire target response in azimuth (Fig. 11A). The wire nearest the transducer was used. The -6 dB beamwidth was 0.68 mm in azimuth and 0.70 mm in elevation compared to an expected beamwidth of 0.55 mm calculated from Field II simulations. In both cases, there are sidelobes at -13 dB and some clutter below -20 dB.

Fig. 12 contains images of the cyst phantom. Fig. 12A-C show two perpendicular B-scans and a C-scan with the short axis of the cyst in azimuth and Fig. 9D-F show two perpendicular B-scans and a C-scan with the short axis of the cyst in elevation. All images in Fig. 12 are log-compressed and are shown with 30 dB dynamic range. Fig. 12A shows the cyst in cross-section.

The cyst is not perfectly circular because of mechanical compression of the phantom to prevent motion during the data acquisition process. In the elevational B-scan and C-scan, the cylindrical cyst appears as a rectangle. Fig. 12D-F show the cyst with short axis in elevation. Although some clutter is present, the cyst is visible in all images.

V. Discussion and Conclusions

We have experimentally verified the feasibility of a row-column device for 3-D imaging using a 5 MHz 256×256 2-D array. The array was made using a modified 1–3 composite fabrication process. Our experimental results indicate the feasibility of 3-D imaging using a row-column transducer array with a reduced fabrication complexity and a decreased number of channels compared to a fully sampled 2-D array of comparable size. Rectilinear volumetric scans with a wide field of view close to the transducer could prove more useful for abdominal, breast, or vascular imaging when overlying bone is absent from the field of view. The impedance and pulse-echo measurements showed good agreement between simulation and experiment. Using this row-column addressing scheme, the high electrical impedance associated with small elements in a typical 2-D array was avoided. The small differences may be partially due to additional parasitic cable capacitance not included in the KLM model. In simulation, the center frequency was 5.2 MHz with a -6 dB fractional bandwidth of 67 %. Experimentally, the spectrum of the pulse has a center frequency of 5.3 MHz and a -6 dB fractional bandwidth of 53 %. The lower experimental bandwidth is believed to be a result of variation in bond thickness since it is quite difficult to achieve highly uniform pressure over a $40 \text{ mm} \times 40 \text{ mm}$ area. As a result, this would lead to bond lines with varying thickness across the array. Using the wire target phantom, the -6 dB beamwidth was 0.68 mm in azimuth and 0.70 mm in elevation compared to an expected beamwidth of 0.55 mm calculated from Field II simulations. In both cases, there are sidelobes at -13 dB and some clutter below -20 dB. Sidelobes from the wire targets and clutter in the anechoic cyst regions are present, which may be due to the variability of element-to-element performance in terms of sensitivity and bandwidth. Nevertheless we are encouraged by these initial results. In theory, real-time 3-D rectilinear imaging is possible if enough system channels and parallel beamformers are available. In this paper, we acquired data and beamformed synthetically due to a limited channel count. Data acquisition takes about 2 hours in the current implementation. Future work includes reducing clutter by other post-processing and filtering techniques and obtaining 3-D images of in-vitro and excised tissue experiments.

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Fig. 1. Row-column addressing scheme. The transmit side consists of 8 horizontal electrodes. The receive side consists of 8 vertical electrodes which serve as ground in transmit. Transmit beamforming is done in the vertical or elevational direction while receive beamforming is done horizontally or azimuthally.

Fig 2.
Transmit and Receive Beamforming for 3-D rectilinear imaging

Fig 3.

a) Modified Tx/Rx circuit where entire row acts as a single transmit element and an entire column acts as a single receiver and b) Circuit diagram used to verify performance of the modified Tx/Rx circuit using a single element transducer.

Fig. 4. Simulated beamplots of the row-column transducer A) 3-D beamplot and B) Contour plot with lines at -10, -20, -30, -40 and -50 dB.

Fig. 5.
Two single layer flexible circuits attached to perpendicular sides of the transducer.

Fig. 6.
Photograph of the finished 256×256 2-D array transducer

Fig 7.

Experimental pulse-echo using modified Tx/Rx circuit and a 5 MHz Panametrics piston transducer. The solid line is the echo received from a glass plate using a standard Tx/Rx channel. The dashed line is the echo where one channel was used as a transmitter only and another channel is used in receive only.

Fig. 8. Experimental (solid lines) and simulated (dashed lines) electrical impedance a) magnitude plots and b) phase angle plots

Fig. 9. Experimental (solid lines) and simulated (dashed lines) a) pulse and spectra of the row-column transducer. The pulse plots were zoomed in to clearly see the two pulses.

Fig 10.

Experimental axial wire target images. A-C are acquired with short axis of the wires positioned in azimuth direction and D-E are acquired with short axis of the wires positioned in elevation direction. All images are log-compressed and shown with 20 dB dynamic range

Fig 11. Lateral wire target response in azimuth direction (A) and in elevational direction (B). The dashed lines is a Field II simulated beamplot and the solid line is the experimental beamplot

Fig 12. Experimental 8 mm anechoic cyst images. A-C are acquired with short axis of the cyst positioned in azimuth direction and D-E are acquired with short axis of the cyst positioned in elevation direction. All images are log-compressed and shown with 30 dB dynamic range