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A 7.5 MHz Dual-Layer Transducer Array for 3-D Rectilinear Imaging

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Abstract

The difficulties associated with fabrication and interconnection have limited the development of 2-D ultrasound transducer arrays with a large number of elements (>5000). In previous work, we described a 5 MHz center frequency PZT-P[VDF-TrFE] dual-layer transducer, which used 2 perpendicular 1-D arrays for 3-D rectilinear imaging. This design substantially reduces the channel count as well as fabrication complexity, which makes 3-D imaging more realizable. Higher frequencies (>5MHz) are more commonly used in clinical for imaging targets near transducers such as the breast, carotid, and musculoskeletal. In this paper, we present a 7.5 MHz dual-layer transducer array for 3-D rectilinear imaging. A modified acoustic stack model was designed and fabricated. PZT elements were sub-diced to eliminate lateral coupling. This sub-dicing process made the PZT into a 2–2 composite material, which could help improve transducer sensitivity and bandwidth. Full synthetic aperture 3-D data sets were acquired by interfacing the transducer with a Verasonics data acquisition system (VDAS). Offline 3-D beamforming was then performed to obtain volumes of a multi-wire phantom and a cyst phantom. The generalized coherence factor (GCF) was applied to improve the contrast of cyst images. The measured –6 dB fractional bandwidth of the transducer was 71% with a center frequency of 7.5 MHz. The measured lateral beamwidths were 0.521 mm and 0.482 mm in azimuth and elevation respectively, compared with a simulated beamwidth of 0.43 mm.

I. Introduction

There are several clinical applications of 3-D ultrasound at present. These applications include identifying ulcers on carotid plaque surface [1], evaluating vessel wall volume (VWV), which is a 3-D measurement of the carotid artery intima and media [2], guidance of autonomous robotic breast biopsy [3], detection of breast tumor bed displacement [4], and monitoring the early stages of fracture healing in the musculoskeletal system [5]. 3-D ultrasound can also be used to perform 3-D photoacoustic and acoustic imaging [6].

Several research efforts to fabricate fully-sampled 2-D arrays using piezoceramics such as lead zirconate titanate (PZT) and 1–3 composites have been reported [7–10]. The center frequencies of these arrays basically range from 2 MHz to 5 MHz. Capacitive micromachined ultrasonic transducers (CMUTs) are also an attractive alternative by using standard silicon integrated circuit technology and the potential for electronic integration. Oralkan *et al.* proposed a 2-D CMUT array of a 4.37 MHz center frequency in [11]. Philips developed a fully-sampled 2-D phased array of 1–6 MHz frequency range with 9,212 elements [12]. However, due to the difficulties in fabricating the high densely populated arrays and providing individual electrical connections to each element, most of the 2-D arrays have less than 10,000 elements and operate at frequencies lower than 6 MHz. Light *et al.* presented catheter-based 2-D transducer arrays operating at frequencies up to 10 MHz with small amount of elements for intravascular imaging [13]. In addition, there are also challenges in acquiring and processing data from a large number of channels. Sparse arrays were designed for 3-D rectilinear imaging focused on suppressing clutter [14–16]. Other potential solutions include a crossed-electrode scheme using a hemispherically shaped array

to scan a pyramidal volume [17] and our previous research on a row-column addressing technique to simplify interconnections of a 4×4 cm² 2-D transducer array [18].

Dual-layer or multilayer transducers have been proposed for diagnostic applications such as acoustic bladder volume assessment [19], simultaneous B-mode and Doppler duplex imaging [20], and harmonic imaging [21]. We previously proposed a dual-layer transducer for 3-D imaging [22]. This transducer contained one PZT layer for transmit and one separate P[VDF-TrFE] copolymer layer, closer to the targets, for receive. Each layer was a 1-D square-shaped array composed of 256 parallel elongated elements. The layers were oriented with transmit and receive elements perpendicular to each other. The dual-layer design enabled us to optimize materials separately for different layers and also isolate transmit and receive electronics.

This paper describes the design, fabrication, test, and imaging experiments of a 7.5 MHz dual-layer transducer with 256 PZT elements and 256 P[VDF-TrFE] copolymer elements. As another realization of the aforementioned dual-layer method, this transducer used a thinner layer of PZT, 125 μ m, which required a re-design of the acoustic stack, and refinement of the fabrication process. A 4×4 cm² prototype was developed to demonstrate the feasibility of the transducer. The generalized coherence factor (GCF) was also applied to improve the image quality furthermore. The ultimate goal of this work is to develop 3-D imaging systems in the frequency range of 7.5–12.5 MHz. These systems are analogous to today's linear array systems which are used for imaging breast, carotid, and musculoskeletal. We aim to have a system with image quality comparable to a system with a fully-sampled 2-D array, using a more practical transducer design combines with suitable signal processing algorithms.

II. Methods

a. Simulated beamplots

To evaluate the theoretical imaging performance, simulated beamplots were acquired using Field II [23]. The transmit aperture was a 1-D array with an azimuthal element pitch of 150 μ m and an elevational height of 38.4 mm. The receive aperture had an elevational element pitch of 150 μ m and an azimuthal length of 38.4 mm. A Gaussian pulse with a center frequency of 7.5 MHz and 50% fractional bandwidth was used. For the beamplot, a 128-element subaperture was used in both transmit and receive and focused on-axis to $(x,y,z) = (0,0,30)$ mm. Figure 1 shows the simulated beamplots of the dual-layer transducer. The lines in Figure 1B are at -10, -20, -30, -40, and -50 dB. The -6 dB beamwidth is 0.43 mm, and the highest clutter levels, around -30 to -40 dB, are seen along the azimuth and elevation axes. The clutter levels drop off dramatically in regions away from the principal azimuth and elevation axes.

b. Design and fabrication

Figure 2 shows the acoustic stack of the dual-layer transducer array schematically. A backing with acoustic impedance of 13 MRayls was used to minimize reverberation between different layers. This backing was produced using 85% tungsten powder (Atlantic Equipment Engineers, Bergenfield, NJ) and 15% Epo-Tek 301 epoxy (Epoxy Technology, Billerica, MA), by weight. 85% of the tungsten particles were of 10 μ m diameter, while the remaining 15% were of 1 μ m. The mixture was then centrifuged at 3000 revolutions per minute (rpm) in a Beckman-Coulter Allaeagra 6 centrifuge (Fullerton, CA). After lapping, one side of the backing was sputtered with 500 angstroms of chrome and 1000 angstroms of gold to provide a ground plane for all PZT elements.

A $40 \times 40 \text{ mm}^2$ active area size was chosen by considering the typical field-of-view used in clinical settings. We also chose a pitch of $150 \mu\text{m}$ for the array to keep the elements spaced less than one wavelength to avoid grating lobes. As a result, each layer had 256 elements. A $40 \times 40 \text{ mm}^2$ wafer of gold-plated $300 \mu\text{m}$ thick PZT-5H was bonded to the sputtered surface of the backing. A DAD321 automatic dicing saw (DISCO Corporation, Tokyo, Japan) with a $28 \mu\text{m}$ width diamond blade (DISCO Corporation, Tokyo, Japan) was first used to dice the PZT wafer into 256 elongated posts at a $150 \mu\text{m}$ pitch. These first cuts were filled with Epo-Tek 301 epoxy. After curing at room temperature in a dry environment for 48 hours, PZT along with the excess epoxy was lapped down to the desired thickness of $125 \mu\text{m}$ and sputtered with chrome/gold electrode on top. A second set of cuts were made by aligning the dicing saw to the center of the ceramic posts created by the first set. All cuts were made through the PZT piece and about $50 \mu\text{m}$ into the backing. After the second dicing process, we created the elements for a 2–2 composite array with a ceramic volume fraction around 70%. This sub-dicing strategy was used to reduce the width-to-height ratio of ceramic posts to around 0.32, which exceeded the requirement for low lateral coupling [24]. Without sub-dicing, the determined transducer specifications would make the width-to-height ratio close to 1, which would translate to high lateral coupling.

A prototype $25 \mu\text{m}$ thick flexible circuit (Microconnex, Snoqualmie, WA, Flex 1 in Figure 1) was then bonded to one edge of the PZT array using nonconductive epoxy, with fine alignment between PZT elements and copper traces of the flex to create electrical connections. The flexible circuits were made of $25 \mu\text{m}$ thick polyimide with $4 \mu\text{m}$ thick copper traces printed on one side. A $40 \times 40 \text{ mm}^2$ sheet of P(VDF-TrFE) copolymer was bonded to the other flexible circuit (Flex 2 in Figure 1) which was identical to Flex 1. A single P(VDF-TrFE) copolymer element was $75 \mu\text{m}$ wide and 40 mm long, defined by the size of copper traces on the flexible circuit. This copolymer/flex module was then bonded to the top of the PZT so that the PZT and copolymer elements were perpendicular to each other. A $3 \mu\text{m}$ thick parylene layer was sputtered on PZT to electrically isolate the elements of the two layers. The copolymer combines with the two flex circuits to serve as a simple matching layer for the PZT transmit layer. In all bonding steps, the applied pressure was approximately 100 psi.

Samtec connectors (Samtec USA, New Albany, IN) were soldered onto both flexes to serve as the interface between the transducer and printed circuit boards. A photo of the finished prototype transducer is shown in Figure 3.

c. Data acquisition

After performing electrical impedance and pulse-echo measurements, the dual-layer transducer array was interfaced with a 4-board Verasonics data acquisition system, VDAS (Verasonics, Redmond, WA) using custom-printed circuit boards. This system allows users to control imaging parameters such as aperture size, transmit frequency, filtering, and time-gain compensation. The VDAS can provide 256 transmit channels and 128 receive channels simultaneously, which reduces the time length of data collection and the complexity of operation significantly.

Figure 4 shows a schematic of transmit and receive elements grouping for synthetic data acquisition. During acquisition, each of the TX1 elements were connected to 128 individual channels of the VDAS configured to operate in transmit mode, while RX1 elements were connected to the remaining 128 system channels in receive mode. Data from each receive channel was collected 100 times with a 36 MHz sampling frequency, and averaged to minimize the effects of random noise. Next, the RX2 group was connected manually to the same set of receive channels, completing collection from all 256 receive element. The above

process was repeated for TX2 group, and all transmit/receive element combinations were acquired.

We acquired 3-D volumetric data of a home-made $70 \times 70 \times 70 \text{ mm}^3$ gelatin phantom containing 5 pairs of $100 \mu\text{m}$ diameter nitinol wire targets with axial separation of 0.5, 1, 2, 3, and 4 mm, to evaluate the spatial resolution. The bottom wire in each pair was laterally shifted by 1 mm with respect to the top wire. This background material of the wire phantom consisted of 400 g deionized (DI) water, 36.79 g n-propanol, 0.238 g formaldehyde, and 24.02 g gelatin. These ingredients and quantities were based on recipes given in [25]. The second phantom had an 8 mm diameter cylindrical anechoic cyst located at a depth of 30 mm from the transducer face. The background of this cyst used the same ingredients as the wire target phantom plus 3.89 g of graphite powder to provide scattering.

d. beamforming, signal processing, and display

The acquired data was imported into Matlab (Mathworks, Natick, MA) for offline 3-D delay-and-sum (DAS) beamforming, signal processing, and image display. RF data was filtered with a 64-tap FIR bandpass filter with frequency range of 4.8 – 10 MHz. Then beamforming with dynamic transmit and receive focusing of 1 mm increments was done with a constant sub-aperture size of 128 elements, or 19.2 mm. A 3-D volume was acquired by selecting the appropriate transmit sub-apertures in azimuth and receive sub-apertures in elevation to focus a beam directly ahead. The dimensions of the acquired volume were $38.4 (\text{azimuth}) \times 38.4 (\text{elevation}) \times 44.5 (\text{axial}) \text{ mm}^3$. The generalized coherence factor (GCF) was calculated and used as weighting factor for the reconstructed images of the cyst phantom to improve contrast of the images [26]. After 3-D beamforming, envelope detection was done using Hilbert transform. Images were then log-compressed and displayed with proper dynamic ranges. Azimuth and elevation B-scans are displayed along with C-scans which are parallel to the transducer face.

III. Experimental results

a. Impedance measurements

Experimental electrical impedance measurements of the transducer were taken using an Agilent 4294A impedance analyzer (Santa Clara, CA). Figure 5 shows the electrical impedance results of both layers by simulation using 1-D KLM modeling software (PiezoCAD, Sonic Concepts, Woodinville, WA) [27], and by experiments. For PZT, the simulated impedance magnitude was 20Ω at a series resonance frequency of 11.5 MHz while the experimental impedance curve showed a series resonance of 45Ω at 11.4 MHz. The phase plots peak at 15.4 MHz for the KLM simulation and at 14.8 MHz in the experimental case. For P(VDF-TrFE) copolymer, the impedance magnitude was 806Ω at 11.4 MHz in simulation, and the measured impedance magnitude was 1.11 k Ω . No resonance peaks are seen in the impedance magnitudes, and the phase remains near 83° to 86° .

b. Pulse-echo measurements

Pulse-echo measurements of the transducer were made in a water tank using a Panametrics 5900PR pulser/receiver (Waltham, MA) with an aluminum plate reflector. To mimic imaging conditions, the excitation pulse was applied to a PZT element and a copolymer element was used as the receiver. Figure 6 shows the simulated and experimental time and frequency responses of the pulse-echo signals. In simulation, the center frequency was 8.8 MHz with a -6 dB fractional bandwidth of 53%, compared to a 7.5 MHz center frequency with a -6 dB fractional bandwidth of 71% in experiment. Low amplitude reverberations

after the pulse peak are seen in both the simulation and experimental pulses in the time domain.

c. 3-D imaging-multiwire phantom

Figure 7(a)–(c) show the azimuth B-scan, elevation B-scan, and C-scan respectively when the short axis of wire targets were in elevation. The azimuth B-scan (Figure 7(a)) shows the pair of wires with 0.5 mm axial separation, and the two wires are discernible. The C-scan, taken at a depth of 35 mm, is parallel to the transducer face. Figures 7(d)–(f) show images of the same phantom with short axis of wire targets in azimuth. The pair of wires with 0.5 mm axial separation is also discernible in the elevation B-scan shown in Figure 7(e). Figure 7(f) shows a C-scan which has been tilted to encompass the entire length of the wires, when the wires are not perfectly parallel to the transducer face because of mechanical positioning. All images are log-compressed and shown on a 20 dB dynamic range.

Figure 8 shows the lateral wire target responses in azimuth (Figure 8A) and in elevation (Figure 8B). In both cases, the wire closest to the transducer was used for measurement. The -6 dB beamwidth in azimuth was 0.521 mm and 0.482 mm in elevation compared to a simulated beamwidth of 0.43 mm in both directions. Sidelobes above -15 dB and some clutter around -20 dB were present in both figures, and more severe clutter was observed in azimuth than in elevation.

d. 3-D imaging-cyst phantom

Figure 9 contains volumetric images of a phantom with an 8 mm diameter cylindrical cyst at 30 mm depth. GCF was applied to minimize the degradation of the images due to incoherence. Figure 9(a)–(c) show the azimuth B-scan, elevation B-scan and C-scan with the short axis of the cyst in elevation. Figures 9(d)–(f) show the two perpendicular B-scans and the C-scan with short axis of the cyst in azimuth. Figure 9(g)–(i) show the scan images with the cyst being positioned diagonally above the transducer surface. Although some clutter is present, the cyst is visible in all images. All images are log-compressed and shown with 50 dB dynamic range. Comparison of contrast-to-noise (CNR) ratio of volumes before and after GCF processing is listed in Table 1. Improved image contrast was achieved by applying GCF according to the CNR values.

IV. Conclusions and discussions

In this paper, we described the design, fabrication, test and imaging of a 7.5 MHz frequency dual-layer transducer array using PZT and P[VDF-TrFE] copolymer for transmit and receive, respectively. Compared to the initial 5 MHz transducer, fabrication of higher frequency dual-layer transducers requires a new design and additional fabrication steps. Design constraints such as element aspect ratio and number of elements became more stringent as the frequency increases. We addressed these considerations in this paper and described methods to design these types of transducers for 7.5 MHz. Experimental measurements of the transducer showed good agreement with simulation results. By imaging the multi-wire phantom, lateral resolution was improved due to a higher center frequency. In order to improve image quality further, we applied GCF on the cyst phantom images which contained speckles from scatterers in the background. The contrast of the images was enhanced after being GCF processed. By combining GCF with a higher center frequency dual-layer transducer, we achieved better image quality in terms of resolution and contrast.

Figure 8 shows that clutter levels were higher in the azimuth direction than in the elevation direction. The difference in clutter levels between the two directions was also reflected in

quality of images. In cyst phantom images, when the long-axis of the cyst was in the azimuth direction, clutter contribution from the azimuth sidelobes was minimized or not even present. This result is consistent with the cyst images which show the highest CNR when the long axis of the cyst was in the azimuth direction (Figure 9(a)–(c)). When the long-axis of the cyst was in the elevation direction, clutter contributions from the azimuth sidelobes were greater resulting in a lower CNR. CNR values were lowest when the cyst lied along the diagonal (Figure 9(g)–(i)). This is expected since clutter contributions from both principal axes were present.

The unwanted clutter was likely due to variation of element-to-element performance from the difficulty of achieving uniform pressure over $4 \times 4 \text{ cm}^2$ area while bonding. In addition, because the dual-layer design allows only one-way focusing in each direction, the transducer will have lower image quality compared to fully-sampled 2-D transducer arrays which are capable of two-way focusing. As a result, signal processing algorithms are required to improve image quality. We took advantage of GCF to improve the contrast of the images in this paper. Future work will focus on investigating other signal processing approaches to enhance image quality further.

Our future work will also involve developing dual-layer transducers of even higher center frequencies ($> 10 \text{ MHz}$) based on 1–3 composite materials for the transmit layer. Additionally, due to the flexibility of the copolymer material, we plan to develop curved dual-layer transducers for 3-D transrectal prostate imaging. We will also investigate modifying the VDAS platform for real-time 3-D imaging.

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Figure 1.
Simulated beamplots of the dual-layer transducer.

Figure 2.
Acoustic stack of the dual-layer transducer array.

Figure 3.
Photo of the prototype dual-layer transducer.

Figure 4.
Schematic of transmit and receive elements grouping.

Figure 5.
Simulated and experimental impedance measurements of PZT and P[VDF-TrFE] layers.

Figure 6.
 Simulated and experimental pulse and spectra of the dual-layer transducer.

Figure 7.
Experimental images of multi-wire. All images are log-compressed and shown with 20 dB dynamic range.

Figure 8.
Azimuthal and elevational lateral wire target responses.

Figure 9.
Experimental images of cyst phantom with GCF applied. All images are log-compressed and shown with 50 dB dynamic range.

Table 1

Comparison of CNR of volumes before and after GCF processing

Cyst orientation	CNR without GCF	CNR with GCF
Short axis in elevation	1.84	3.47
Short axis in azimuth	1.72	2.36
Diagonal	1.44	1.76