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Micromachining Techniques in Developing High-Frequency Piezoelectric Composite Ultrasonic Array Transducers

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Abstract

Several micromachining techniques for the fabrication of high-frequency piezoelectric composite ultrasonic array transducers are described in this paper. A variety of different techniques are used in patterning the active piezoelectric material, attaching backing material to the transducer, and assembling an electronic interconnection board for transmission and reception from the array. To establish the feasibility of the process flow, a hybrid test ultrasound array transducer consisting of a 2-D array having an 8×8 element pattern and a 5-element annular array was designed, fabricated, and assessed. The arrays are designed for a center frequency of ~ 60 MHz. The 2-D array elements are $105 \times 105 \mu\text{m}$ in size with $5\text{-}\mu\text{m}$ kerfs between elements. The annular array surrounds the square 2-D array and provides the option of transmitting from the annular array and receiving with the 2-D array. Each annular array element has an area of 0.71 mm^2 with a $16\text{-}\mu\text{m}$ kerf between elements. The active piezoelectric material is $(1 - x) \text{Pb}(\text{Mg}_{1/3}\text{Nb}_{2/3})\text{O}_3\text{-}x\text{PbTiO}_3$ (PMN-PT)/epoxy 1–3 composite with a PMN-PT pillar lateral dimension of $8 \mu\text{m}$ and an average gap width of $\sim 4 \mu\text{m}$, which was produced by deep reactive ion etching (DRIE) dry etching techniques. A novel electric interconnection strategy for high-density, small-size array elements was proposed. After assembly, the array transducer was tested and characterized. The capacitance, pulse–echo responses, and crosstalk were measured for each array element. The desired center frequency of ~ 60 MHz was achieved and the -6-dB bandwidth of the received signal was $\sim 50\%$. At the center frequency, the crosstalk between adjacent 2-D array elements was about -33 dB . The techniques described herein can be used to build larger arrays containing smaller elements.

I. Introduction

Ultrasound is used extensively in medical imaging and for the detection of defects in engineering materials. The need for higher image resolution in medical imaging for specific applications has attracted much attention in the development of high-frequency (HF) ultrasound transducers operating at 20 MHz and above, especially high-frequency array

transducers [1]. These applications include ophthalmology [2]–[4], dermatology [5]–[7], *in vivo* gastrointestinal (GI) imaging [8], [9], and small animal studies [10], [11], for which imaging subsurface structures with microscopic resolution is required. Compared with the fixed depth-of-field and limited frame rate single-element transducer imaging system, high-frequency array-transducer-based systems are desirable for their beam steering, beamforming (on reception), and higher frame rates. Design and characterization of high-frequency single-element ultrasonic transducers have been thoroughly studied and optimized [12], [13], and there are a variety of different production methods and piezoelectric materials available for the production of single-element transducers [14]–[16]. However, there are few alternatives for the development of high-frequency ultrasonic array transducers, which face such challenges as limitations in fabrication technology and equipment, the absence of suitable high-quality piezoelectric materials and the absence of convenient circuit boards to read out the signals of each element during the testing phase. At the present time, the only commercial array-based imaging system above 20 MHz is from VisualSonics Inc. (Toronto, Canada); it is targeted for high-resolution imaging of small animals [17].

The challenges for HF array transducer manufacturing are mainly related to the small dimensions, narrow pitches, and large number of array elements, which result in major problems in selecting 1) the piezoelectric material, 2) an appropriate technology for fabricating these miniature piezoelectric elements, and 3) a strategy for electrically interconnecting these small array elements with narrow pitches. In addressing these challenges, a few innovative approaches have been developed.

The most crucial component of an ultrasound transducer is the piezoelectric element. There are many piezoelectric materials available, but a high electromechanical coupling coefficient and good acoustic matching on an element-by-element basis are necessities. Also, low crosstalk between nearby elements is particularly important for arrays containing small elements with narrow pitches. Because of their unique structure, piezoelectric composite materials have many attractive characteristics for ultrasonic transducer applications in general, and high-frequency array transducers in particular. The 1–3 piezoelectric composite selected for the test system described herein offers a large number of design parameters including the piezoelectric material pillar dimension, epoxy selection, and volume fraction of the selected piezoelectric material [relaxor-based single-crystal (1 – x)Pb(Mg_{1/3}Nb_{2/3})O₃– x PbTiO₃, (PMN-PT)], which can easily be optimized to satisfy performance requirements. In a 1–3 composite, piezoelectric pillars are embedded in soft epoxy matrices, and the lateral clamping of the active piezoelectric element is greatly decreased. This gives rise to a larger piezoelectric coupling coefficient (k_t) than that which is available with the bulk material operating in the plate mode ($k_t \sim 0.58$) [18]. Furthermore, this special structure can significantly decrease inter-element crosstalk, which is very critical for a high-frequency array transducer with small-size elements for the reasons previously noted. In addition, with the presence of epoxy, the calculated acoustic impedance [19], [20] of the composite is much lower (~ 16 MRayl) than the bulk PMN-PT single crystal (~ 37.8 MRayl) used in this study. The latter value corrects the results of [21] for a mass density of 8.2 g/cm^3 . The composite improves acoustic impedance matching between the transducer and human tissue and facilitates wideband performance for increased axial resolution.

Typically, 1–3 composite is formed of repeated square pillars of piezoelectric ceramic or single crystal. The lateral periodicity of the pillars in the composite material introduces potential additional spurious resonances. To avoid the lateral modes around the designed center frequency, the 1–3 composite material must have pillars and kerfs with very small dimensions [22], [23] so that the lateral mode is at a frequency much higher than the upper band edge of the ultrasound transmission and nominally twice the center frequency. For example, at 60 MHz, the pillar size should be less than 10 μm and a kerf width of $\sim 5 \mu\text{m}$ or less should be selected. Physical models and optimal design have been reported to tailor the properties of composite piezoelectric materials for medical ultrasonic transducer applications [24], [25]. The shape and layout of piezoelectric pillars can be adjusted to avoid lateral modes in the passband of interest and/or reduce lateral mode amplitude [26]. Other disadvantages, such as lower permittivity and higher fabrication complexity, can be overcome by applying impedance matching in the system and selecting suitable fabrication methods, respectively.

The traditional approach for composite construction is the mechanical dice-and-fill method [27]. The width of the dicing blade limits the kerf width to 10 to 15 μm , which is appropriate for transducers operating below ~ 20 MHz. Other alternative methods include the interdigital bonding technique [28], stacked plates or lamination techniques [29], fiber processing [30], and laser machining [31], [32]. A review of these methodologies is provided by Safari [33]. Micromachining processes inherited from the microelectronics industry have been successfully applied in a wide variety of scientific and technological areas such as biology, the life sciences, the automobile industry, and medical device manufacturing. This technology has been also fruitfully utilized in ultrasonics. For example, micromachining is used in the construction of capacitive micromachined ultrasonic transducers (CMUTs) which are extensively researched for medical imaging [34]–[36]. Deep reactive ion etching (DRIE) etching processes are used for etching PZT [37]–[39] and PMN-PT [40]. Photolithography has been used to fabricate flexible circuits and pattern electrodes for array transducers. Similar microfabrication techniques involving chlorine-based plasma ion beams can be used to pattern bulk PMN-PT to form a composite [41], [42]. This is the technique that is adopted in the current study. Although recent studies [26], [43] indicated that different pillar shapes or randomly placed pillars could break the periodicity in the composite material and thus dramatically suppress the lateral modes, it is a formidable task for traditional fabrication methods to fabricate such composites. Fabricating composites with exotic shapes or arrangements of the pillars, however, is not a concern for micromachining techniques at high frequencies, and this will be the focus of a future paper.

In addition to the challenge of producing piezoelectric composite materials for HF ultrasound array transducers, electrical interconnection is another difficult issue that must be solved in fabricating HF array transducers, which have a large number of elements with narrow pitches. In the past, a variety of interconnection schemes has been employed, but most can only be used when the space between elements is large or if the pitch is large. For example, the use of flexible circuits for 2-D arrays has a practical limit on pitch size of $\sim 150 \mu\text{m}$ [44]. Current electric interconnection methods for array transducers include wire bonding [45], [46] for annular arrays and for linear arrays, single-layer flexible circuits [47] for linear arrays, and multilayer ceramic [48] and flexible circuits [45], [49] for 2-

dimensional arrays. With the wire bonding method, fine gold or aluminum wires must be bonded onto each element one by one; this limits the utility of the process to larger elements. Single or multilayer flexible circuits are directly attached to the array elements to expand electric connections. The backing material is then added on the other side of the circuits. They are widely used in low-frequency commercial array transducers. However, the acoustic impedance mismatch between flexible circuits and the backing material sometimes may not be tolerable for high-frequency, wideband, ultrasound transducers. Electrical connection to each element is preferentially made on the back side of the transducer, which simplifies the top radiating surface. The most suitable choice for a high-frequency transducer is the direct attachment of the backing material [42] to the backside of the transducer. Electrical interconnection must reach the back electrode of the element through the backing material. Alternatively, conductive backing material may serve as a via for electric connection. As a result, insulating material must be placed between the elements to separate the conductive backing material.

To overcome these problems, methods based on micro-machining and microelectromechanical systems (MEMS) technology have been systematically studied and employed for the fabrication of high-frequency ultrasonic array transducers. In this study, PMN-PT/epoxy 1–3 composite is produced by micromachining processes combining photolithography and dry etching techniques. A conductive backing material is directly added to each array element, separated by an insulating mold fabricated using photolithography. In addition, a new electric interconnection strategy is implemented that can be applied to many types of array transducers with fine elements and narrow pitches. The new methodology makes use of microfabrication techniques, similar to interposer fabrication in silicon technology, to construct an electric interconnection board (EIB), which serves as the interface between the array elements and the electric system. A hybrid test array transducer was designed, fabricated, and evaluated to demonstrate these fabrication techniques.

II. Design of a Test PMN-PT Composite Ultrasound Array

In this paper, micromachining techniques for the fabrication of high-frequency, wideband array transducers are discussed. To establish the feasibility of the techniques, we constructed a test PMN-PT/epoxy 1–3 composite array transducer, including an 8×8 element 2-D array and a 5-element annular array. The annular array provides the option to transmit on the annular array and receive on the planar array with appropriate electronics. Fig. 1 shows the layout of the transducer for a center frequency of 60 MHz. The red square in the middle designates the area of the planar array and the two green concentric circles delineate the 5-ring annular array. The outer diameter of the annular array is 2.6 mm, and the active 2-D array is $875 \times 875 \mu\text{m}$ in size. Detailed design parameters are listed in Table I.

As mentioned earlier, composite piezoelectric materials have many advantages such as a high electromechanical coupling coefficient and low acoustic impedance. However, there are lateral mode resonances because of the repeated pillars of the composite, which introduce artifacts in the signal return and consume some of the input power. A diagram of the composite material is provided in Fig. 2. The kerf width between pillars in the

composites is on average about $4\text{ }\mu\text{m}$ and the lateral dimension of the pillars is $8 \times 8\text{ }\mu\text{m}$. For this design with 49% volume fraction, the first lateral mode frequency $f_{L1} = 123\text{ MHz}$, which is roughly twice the center frequency of the transducer.

The thickness of the composite was set to $24\text{ }\mu\text{m}$ to accommodate a center frequency of 60 MHz . The element size in the planar array was large ($105 \times 105\text{ }\mu\text{m}$) for test purposes; this allowed issues related to constructing a connection circuit board and the deposition of viscous conductive epoxy backing material behind each element to be more readily examined and resolved. The individual $105\text{-}\mu\text{m}$ elements have very narrow -6-dB beam widths at 60 MHz ($\pm 6^\circ$ relative to vertical), which decreases lateral resolution but increases signal-to-noise ratio in the near field. In addition, the large element pitch ($110\text{ }\mu\text{m}$) of the test system generates strong grating lobes which limit beam steering in the near field. Smaller elements in greater numbers are required for medical imaging applications. This is discussed in greater detail in Section V. Although the current design does not satisfy the criterion of $\lambda/2$ pitch for phased arrays, the bigger elements have smaller electric impedance compared with the several thousand ohms for the smaller elements and it is still a reasonable test device given the many manufacturing challenges. Therefore, this design has been implemented to evaluate fabrication techniques for the construction of high-frequency ultrasonic array transducers.

The transducer architecture is illustrated in Fig. 3. The transducer stack includes a ground electrode, PMN-PT/epoxy 1–3 composite, array electrodes, conductive backing material, and the insulator separating the backing material between array elements. Thin array electrodes ($2000\text{ }\text{\AA}$ thick) are patterned on the back side of the composite and the electrically conducting backing material is directly attached to each array element electrode. An insulating mold constructed using SU-8 is aligned with the kerfs and is used to isolate the backing material behind each element. This architecture guarantees good acoustic impedance matching between piezoelectric components and the backing material, which is critical for high-frequency transducers. An array of pins on the EIB directly attach to the center of the backing material on each element.

III. Fabrication of Ultrasonic Array Transducer Stack

A. Ultrasonic Array Stack Fabrication

As mentioned, in the design described in the previous section, the kerf width between PMN-PT (H.C. Materials, Bolingbrook, IL) pillars in the composite material is as small as $4\text{ }\mu\text{m}$. The traditional dice-and-fill method and other precise machining methods including laser ablation cannot accomplish the task or are time-consuming. In this work, the PMN-PT/epoxy 1–3 composite was produced using microfabrication processes combining photolithography and dry etching techniques. The production of the composite and the array stack have been thoroughly explained in [50]. This method has several advantages, such as batch processing, submicrometer machining precision, and low cost for mass production.

The scanning electron micrographs (SEMs) in Fig. 4 illustrate the sample at different processing stages during pillar structure construction for the 2-D planar array and the annular array. Note that the PMN-PT structures in Fig. 4(E) are not perfect 90° pillars.

Rather, the side wall angle is 87° (where vertical is 90°). The widening of the columns at the bottom is the result of the dry etching process. A substantial effort was made to limit the widening near the base. This included periodic changes in the orientation of the substrate during the dry etching process. The widening evident in Fig. 4 does not have a significant impact on the acoustic impedance of the composite [51], but it does lower the electromechanical coupling coefficient k_t , as discussed in Section IV.

After the dry etching process, the kerfs were filled in with an epoxy, Epo-Tek 301 (Epoxy Technology, Bellerica, MA). Once the epoxy fully cured at 40°C overnight, the surface was lapped down to PMN-PT level. Electrodes for the 8×8 planar 2-D array and the 5 annular array elements were patterned with the aid of photolithography. Separations of $5\text{ }\mu\text{m}$ between 2-D elements and $16\text{ }\mu\text{m}$ between annular array elements were generated. The sample was subsequently flipped over and was lapped down to the desired thickness of $24\text{ }\mu\text{m}$. Another $500\text{ }\text{\AA}/1500\text{ }\text{\AA}$ thick layer of Cr/Au was then sputtered on the transducer arrays as the ground electrode.

Adding backing material directly to the composite has been found to be the choice for high-frequency ultrasound transducers, which guarantees good acoustic impedance matching and the most effective damping for wideband performance. A silver conductive epoxy, E-Solder 3022 (VonRoll Isola Inc., New Haven, CT) was selected as the backing material for the array. SU-8 insulating structures were built along the gaps between elements on the side where the array electrodes were patterned. These structures served as molds for the fill-in of conductive backing material. Fig. 5 illustrates the insulating structures and the filled backing material.

B. Electrical Interconnection Board (EIB)

Because of the highly dense array elements (with a pitch as small as $110\text{ }\mu\text{m}$ for the 2-D array), electric connection of all the array elements is challenging. A similar methodology for electrical interconnection in transducer arrays is presented here [50]. The basic concept entails the use of metal wires/traces (one for each of the array elements) that are fabricated on an EIB. The wires are embedded under an insulating material. The ends of the wires in the middle of the EIB are centered on the individual array elements. The wires emanating from the center are expanded in width to the connection pads at the edge of the EIB. The ends of traces in the middle of the board have a tiny metal pin ($20\text{ }\mu\text{m}$ in diameter) extending above the insulating layer for each planar array element. For the annular array, there are four pins corresponding to each ring element, which guarantees a high-reliability connection. Following an alignment process, the pins/needles are aligned with the array elements and can easily be pressed into the backing material, E-Solder 3022, at the back side of the transducer array. In so doing, an electrical path from each element to the source excitation pulse is established. A schematic drawing of the EIB is presented in Fig. 6.

Fig. 7 shows the layout and the pictures of the fabricated EIB. Fig. 7(A) demonstrates the layout of the EIB. The whole dimension of the EIB is $25 \times 25\text{ mm}$ and the pitch of the connection pads at the edge of the EIB is as large as 0.8 mm , which is spacious enough to solder cables manually. Fig. 7(B) is the surface profile of the middle area of the EIB where nickel pins/needles are located. Fig. 7(C) shows SEMs of the pins/needles. Under the

insulating layer, there are embedded traces to connect the pins/needles with the connection pads. The pins/needles are very uniform, with a height of 25 μm and a diameter of 20 μm . The fabrication process of the EIB has been explained in detail in [42].

C. Array Transducer Assembly and Packaging

After the array transducer stack and the EIB were fabricated, assembly could begin. Fig. 8 depicts the assembly process involving the connection of the transducer to the board. During bonding, each part was held by a vacuum chuck with the backing material side of the transducer facing the side of the EIB on which the pins were located. The two parts were aligned, with each pin of the EIB corresponding to an array element of the transducer. With a moderate pressure, the Ni pins on the EIB were inserted into the backing material behind each element. Non-conductive epoxy was used to further bond the two parts together. Thus, electrical connections were established between each array element and a corresponding separated pad at the edge of the board through the conductive backing material, the nickel pin/needle, and the embedded Cr/Au trace.

After assembly, a layer of parylene (PDS 2010, Speed-line Technologies, Franklin, MA) was deposited on the front of the transducer, which served as the matching layer between the transducer and water to increase system bandwidth. Coaxial cables were soldered onto the connection pads of the EIB, and SMA connectors were attached to ends of each cable for testing only. A plastic tube glued onto the EIB served as a water reservoir during transducer testing. Finally, the entire unit was placed on a copper plate for grounding purposes and secured inside a small aluminum box which served as an RF shield. Fig. 9 provides a view of the completed test system.

As shown in the top and bottom views at high magnification in Fig. 9, the top surface contains only a single grounding wire; the embedded wires can easily be seen in the bottom view through the glass substrate. There is one trace connected to one 2-D array element, whereas along each diagonal, there are four traces that connect to an annular ring element.

IV. Testing and Experimental Results

The electrical impedance of this single-crystal PMN-PT/epoxy 1–3 composite material was measured under free resonator conditions with no mechanical load and no extra materials attached using a precision impedance analyzer (4294A, Agilent Technologies Inc., Santa Clara, CA). To satisfy the previously described requirements, one piece of 1×1 mm composite material with only top and bottom electrodes was prepared for electric impedance measurement. As indicated previously, the electromechanical coupling coefficient k_t in 1–3 composites is enhanced because of the embedding of PMN-PT pillars in the soft passive epoxy matrix. For the thickness mode, the value k_t can then be calculated using the formula [52]

$$k_t^2 = \frac{\pi}{2} \frac{f_r}{f_a} \tan \left(\frac{\pi}{2} \frac{f_a - f_r}{f_a} \right), \quad (1)$$

where f_r is the resonant frequency (i.e., minimum at serial resonance) and f_a is the antiresonant frequency (maximum at parallel resonance). The value of k_t for the PMN-PT composite was experimentally measured using the electrical impedance procedure set forth in the *IEEE Standard on Piezoelectricity* [52]. The electromechanical coupling coefficient of k_t of the composite was measured to be 0.72, which is much higher than that of PMN-PT single crystal (~ 0.58). Although the linear difference between the two values appears to be small, the transducer performance is nonlinearly dependent on k_t . The higher k_t value of 0.72 greatly facilitates the design of high-frequency, ultra-wideband transducers and significantly improves the two-way amplitude response. Moreover, the reduced acoustic impedance of the composite greatly aids in the selection of optimal backing materials for the transducer. The theoretically derived k_t estimate for a PMN-PT composite with perfectly vertical pillars is ~ 0.9 [53]. However, as noted in Section III, the wall angle after the dry etching process is $\sim 87^\circ$, which results in a change in the desired kerf width from 5 μm at the top of the pillar to 2.5 μm at the bottom. This taper serves to significantly reduce the piezoelectric coupling in the thickness mode [54] resulting in reduced k_t .

To determine the uniformity of all the elements, the capacitances of array elements were measured at the frequency of 1 kHz, as shown in Fig. 10, using an LCR meter (QuadTech 1715 LCR Digibridge, TestEquity, LLC., Moorpark, CA). The average value of the capacitances of the 2-D elements is about 5.5 pF and the average value for the annular elements is about 358 pF.

According to the experimental results, the capacitances of the 2-D elements at the edges or corners of the array were slightly smaller than in the middle. The highest value is 5.8 pF and the lowest value is 5.3 pF. This can arise because the alignment of the patterned array electrodes is less than perfect, the side wall of SU-8 structures was not exactly straight, or a small variation in the thickness of the composite layer exists. Because it is likely that all these defects can be improved or avoided in later fabrication, this suggests that the proposed electric interconnection strategy is feasible for high-frequency ultrasonic transducer array fabrication.

The pulse-echo responses of the transducer elements were measured. In the experiment, deionized water was poured into the reservoir and an X-cut quartz target was suspended in the water above the transducer. The water temperature was 20°C. We used a pulser-receiver (Model 5900PR, Panametrics, Waltham, MA) in pulse-echo mode. The -3-dB receiver bandwidth was 200 MHz. The negative-going square pulse used for transmissions was set to 1 μJ , and the -3-dB bandwidth of the transmitted waveform was ~ 120 MHz. The reflected signals were received and digitized at 500 Msamples/s using an 8-bit digital oscilloscope (LC534, Lecroy, Chestnut Ridge, NY). The electrical impedance of the 2-D array elements at 60 MHz was large ($\sim 250 \Omega$) compared with the receiver impedance (50Ω) because of the small size of the elements. In this preliminary test, the transducer was not electrically matched to the receiver. A typical pulse-echo waveform of one 2-D element is displayed in Fig. 11. The -6-dB echo signal length is 23 ns. The measured results show that the central frequency is 63.5 MHz, and the -6-dB bandwidth is about 46.3%. The full spectrum of the echo return is shown on a logarithmic scale in Fig. 12.

A small (−30 dB) subharmonic is present in the spectrum of Fig. 12 as well as a weak (−33 dB) second harmonic. The subharmonic most likely arises from small microbubbles that were present on or near the quartz target at the time of the measurement. Efforts were made to degas the water with a 30-kHz ultrasound degassing system before the measurements. However, no matter how well the gas concentration in the water is decreased by ultrasonication, the gas concentration will immediately begin to increase once the sonification is stopped. During the 90 min following sonification when the setup and measurements took place, it is estimated that the gas concentration recovered to ~27% of the equilibrium level [55]. Microbubbles are a known and common source of subharmonics [56], [57]. The bubbles may also account for the second harmonic, but a mild nonlinearity can also occur in the propagating of a 60 MHz waveform at a distance of ~1.4 mm [58]. This latter effect could also give rise to a small second-harmonic return. Finally, the response of the Panametrics receiver to the transmission of a pulse generates a weak, low-frequency artifact in the observed spectrum. This spectral component was measured separately and is shown in red in the figure. The artifact weakens with increasing time after transmission and is very small at the time delay of the echo. The strongest feature is the peak at 4 MHz (−28 dB), which is far outside of the spectral region of interest. From the biomedical perspective, there will always be gas in the fluids of living tissue diagnosed by ultrasound, and the effects of bubbles and other artifacts are usually digitally filtered out of the data before image formation.

The insertion loss (or the relative pulse–echo sensitivity) of the transducer is the key parameter that affects imaging quality. It is defined as the ratio of the received transducer echo power P_r to the excitation pulse power P_0 delivered into the transducer. The insertion loss (IL) was determined by

$$IL = 10 \log \frac{P_r}{P_0} = 10 \log \frac{V_r^2 / R}{V_0^2 / R} = 20 \log \frac{V_r}{V_0}. \quad (2)$$

Using an experimental setup similar to that reported by Sherar and Foster [59], the insertion loss of one typical element was measured and the minimum insertion loss at the central frequency of ~60 MHz was about −19.5 dB. The crosstalk for adjacent elements was also measured. The level of electrical and acoustical separation between elements was determined by measuring crosstalk. A function generator (AFG2020, Tektronix Inc., Beaverton, OR) was set to burst mode and used to excite a reference element. The voltages on adjacent elements were measured and compared with the reference voltage. This process was repeated at discrete frequencies over the array pass-band. These results are presented in Table II. Crosstalk at the center frequency was < −30 dB, which is adequate for most applications.

These experimental results indicate that the hybrid test array transducer was operating in a satisfactory manner, which demonstrates that the micromachining techniques reported in this paper are promising for the fabrication of high-frequency ultrasound array transducers.

V. Discussion and Conclusions

This paper reports an effort in attempting to develop suitable micromachining techniques for high-frequency ultrasound array transducer fabrication. Using these methods, an ultrasound array transducer of satisfactory performance was designed, fabricated, and tested. Dry etching techniques were used to produce piezoelectric composite materials from ZnO, PZT film, PZT bulk ceramics, and single-crystal PMN-PT. The chemical mechanisms were generally classified into two types: fluorine-based or chlorine-based. Previous work that showed chlorine-based plasma yielded better profile of the structures was confirmed in the present study. The present study further showed that compared with PZT, PMN-PT gave better profiles with a slope angle that could reach 87° .

The capacitances of 2-D array elements and annular array elements were measured. The ratio of the measured values was about 65, which coincided with the area ratio of the two different array elements. For electric interconnection, each nickel pin corresponded to one 2-D array element. However, because the area of the annular array elements was much larger, there were four pins designed for just one ring element. Therefore even if there is one or more pins missing for an element, as long as there is one pin left and connected to the element, the interconnection still can be maintained. This design can guarantee greater reliability in electric interconnection. The conductive backing material was directly attached to the active array elements and insulated by the SU-8 mold structures.

In conclusion, micromachining techniques have been developed and systematically applied in all key steps in the fabrication of a high-frequency ultrasonic array transducer. These include the fabrication of PMN-PT/epoxy 1–3 composite material, patterning of array electrodes, and attachment of conductive backing material. An innovative electric interconnection strategy was also developed. The 60-MHz ultrasonic array transducer designed and fabricated consisting of 64 2-D elements and 5 annular elements yielded satisfactory performance. These results convincingly demonstrate the promising potential of the micromachining techniques for the fabrication of high frequency ultrasonic array transducers.

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Fig. 1.
Design of a test hybrid ultrasound array transducer with 64-element 2-D array and 5-element annular array.

Fig. 2.

A partial (1/4) model of PMN-PT/epoxy 1–3 composite with 2-dimensional electrodes patterned.

Fig. 3.
Cross-section of a PMN-PT composite ultrasound array transducer stack.

Fig. 4. PMN-PT 1–3 composite structures by dry etching. A: Overview of the array transducer; B: close view of photoresist structures after Ni electroplating; C: Ni mask after photoresist stripped off; D: PMN-PT pillars after dry etching with one expanded view picture showing Ni mask residue left on the tops after etching; E: expanded view of pillars shown in D.

Fig. 5.

Conductive backing material attached to the array transducer. A: Overview of the insulating mold; B and C: conductive backing material filled in the insulating mold; D: cross-sectional view of the transducer after backing material was applied.

Fig. 6.

Model of an electric interconnection board (EIB) for the transducer with 2-D planar array elements and annular array elements. Glass is used as the substrate. Metal traces are patterned onto the substrate, and then they are coated with an insulating layer. One pin for each planar array element is placed at the ends of traces located in the middle area of the EIB; four pins are placed under each annular element.

Fig. 7. Design of a test electric interconnection board (EIB) and pictures of the EIB with pins. A: Layout of the EIB; B: surface profile of nickel pins/needles with the height of 25 μm ; C: SEM pictures with close view of nickel pins/needles.

Fig. 8.

Assembly of the array stack with the electric interconnection board (EIB). With alignment, each 2-D array element is corresponding to one pin, and there are two pins for each annular array element.

Fig. 9.

Pictures of a testing array transducer with 8×8 2-D elements and 5 annular elements assembled with an electric interconnection board (EIB).

Fig. 10. Measured capacitance of the 2-dimensional and annular array elements. For the 2-D elements, the elements are numbered from left to right, row by row; for annular elements, they are indexed from inside to outside.

Fig. 11.

Pulse-echo response of a typical 2-D array element. The quartz target was used and positioned 1.5 mm away from the transducer. The center frequency is 63.5 MHz, and the -6-dB bandwidth is 46.3% of the center frequency or 29.4 MHz.

Fig. 12.

Logarithmic spectral display of the pulse echo response shown in Fig. 11. The eight-bit digitizer used for the measurements yields a dynamic range of 48 dB. Spectral structures are present at the second harmonic as well as at the 1/2 sub-harmonic centered near 30 MHz. No lateral modes are evident in the spectrum. The spectrum of the residual signal resulting from the coupling of the transmitter pulse into the receiver system is over plotted in red.

TABLE I

Design Parameters for the Array Transducer.

Array type	2-D array	Annular array
Number of elements	64	5
Area of one element	0.011 mm ²	0.71 mm ²
Kerf width between elements	5 μ m	16 μ m
Center frequency	60 MHz	
Kerf filler	Epo-Tek 301 (Epoxy Technology, Billerica, MA)	
Active material	24 μ m thick PMN-PT/epoxy 1–3 composite (PMN-PT volume fraction of 49%)	

TABLE II

Crosstalk Between Adjacent Elements.

Frequency (MHz)	Average crosstalk (dB)	Range of value relative to average (dB)
60	-33	-39 to -30
80	-27	-33 to -24
100	-25	-29 to -22