This is the Pre-Published Version.

The following publication Wong, C. M., Chen, Y., Luo, H., Dai, J., Lam, K. H., & Chan, H. L. W. (2017). Development of a 20-MHz wide-bandwidth PMN-PT single crystal phased-array ultrasound transducer. Ultrasonics, 73, 181-186 is available at https://doi.org/10.1016/j.ultras.2016.09.012.

Development of 20-MHz PMN-PT Single Crystal Phased-Array Ultrasound Transducer for Biomedical Imaging Applications

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Abstract

In this study, a 20-MHz 64-element phased-array transducer with a one-wavelength pitch is developed using a PMN-30%PT single crystal and double matching scheme. With an optimized fabrication process involving photolithography technique, excellent piezoelectric and electromechanical coupling properties of the single crystal have been demonstrated to be suitable for wide-bandwidth and high-sensitivity phased-array transducer application. A -6 dB bandwidth of 91% and an insertion loss of 29 dB for the 20-MHz 64-element phased-array transducer are achieved. This significantly improved bandwidth and insertion loss suggest great potential to be implemented in high-resolution biomedical imaging, especially for small animal biomedical applications.

Introduction

High-frequency (≥20 MHz) ultrasound array transducers have been widely used in clinical applications such as ophthalmology, dermatology, and intravascular applications[1]-[5]. By increasing the center frequency of the array transducer up to 30 MHz, the array transducer is significant applicable in small animal studies and human tissue imaging[6]. Single-element ultrasound transducers with the ultrahigh frequency of up to 300 MHz[7,8] have also been extensively studied due to the relatively less complexity and limitation compared to the array transducers. Nevertheless, the single-element transducers exclusively rely on the mechanical scanning for acoustic imaging that may affect the image quality. On the other hand, array transducers have several merits, such as high-quality real-time blood flow measurement, high frame rate, clinical convenience, and capability of dynamic focusing[9]. Moreover, when compared with the single-element transducers and linear array transducers, array transducers are extremely capable for forward-looking intravascular ultrasound (FL-IVUS) and forward-looking endoscopic ultrasound imaging (FL-EUS) due to the wider field of view in the far field[10]. High-frequency array transducers with a frequency higher than 20 MHz, however, are not widely reported due to the challenges from the spatial scale, fabrication method, and hardware beam forming.

The limitation of array transducers is especially shown on the intravascular imaging application even though the array transducers have been extensively developed in the recent decade. Usually, elements in an array transducer are separated either in physical format (kerfed arrays) by mechanical dicing or laser micromachining method or in electrode format (kerfless arrays)

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by photolithography. The kerfs between the elements in the kerfed arrays are filled with epoxy to prevent collapsing of the elements. Various fabrication methods have been reported, but the pitch of those studies was usually larger than $\lambda/2$, and therefore, the grating lobe cannot be totally eliminated, resulting in bandwidth diminishment[11], [12]. A 35-MHz phased-array transducer with a $\lambda/2$ (25 μm) pitch was fabricated by using a double-index dicing method by Cannata et al[13]. The array has shown an average -6 dB bandwidth of 58% at a center frequency of 34 MHz. Alternatively, the kerfless array can also be fabricated by using photolithography to separate the elements by electrodes instead of physical separation[10], [14]-[17]. This method is generally used to fabricate an annular array because the circular kerf cannot be made by the dicing saw. Compared to the kerfed array, the kerfless array has relatively less limitations particularly in pattern fabrication with higher reliability. Moreover, the separation between adjacent elements can be narrower than the diced kerf. Thus, the active area of elements is increased and the pitch of array is reduced, respectively. However, the kerfless array would exhibit relatively high crosstalk and long ring-down time due to the signal coupled between adjacent elements. Therefore, the performance of kerfless array is still relatively lower than that of kerfed array unless the undesired factors are overcome in the system design of kerfless array[10], [18].

Besides the array configurations, the properties of piezoelectric materials are also critical for array transducers performance, in which large piezoelectric coefficients and electromechanical coupling factors are crucial for high-sensitive and wide-bandwidth transducers. Among all commercialized piezoelectric materials, a single crystalline piezoelectric material, lead magnesium niobate-lead titanate (PMN-PT), has the highest piezoelectric coefficients and largest electromechanical coupling factors, which is emerging in many new commercialized medical imaging instruments and has been used in wide range of applications that require high sensitivity and small aperture transducer design[19]-[23]. However, the brittle nature of single crystal induces cracks easily during the mechanical dicing process for small pitch design, and this is the bottleneck for such material being used in high-frequency ($\geq$20 MHz) array transducers.

This report demonstrates a feasible fabrication process of a 64-element 20-MHz phased-array transducer with wide-bandwidth and high-sensitivity for biomedical applications. The transducer exhibits excellent performance, which is one of the best among the reported results[4], [24]-[26]. Nevertheless, in this report, the piezoelectric layer was designed with a one-wavelength pitch (75 μm). Although the pitch is not ideally $\lambda/2$ (37.5 μm), high imaging quality could still be acquired with the compensation of wide-bandwidth design[27].

Array design and fabrication method

A PMN-xPT (with x = 0.30) single crystal was used as the piezoelectric layer of array transducer, and the main piezoelectric properties of the PMN-PT single crystal are listed in Table I. The array transducer was assembled by a conductive epoxy backing layer and double matching layers, where the first matching layer was made by alumina powder mixed with epoxy and the second layer was pure epoxy. Figure 1 shows the schematic of designed phased-array transducer, and the acoustic properties of matching layers and backing layer are listed in Table II.

TABLE I. Properties of PMN-30%PT single crystal

Material	PMN-30%PT
Longitude velocity	4200 ms^{-1}
Density	7500 kg/m^3
Acoustic impedance	31.5 MRayls
Piezoelectric constant ($\boldsymbol{d_{33}}$)	1400 pC/N
Clamped dielectric constant ($\boldsymbol{\varepsilon_{33}^{s}}$)	1400
Dielectric constant ($\boldsymbol{\varepsilon_{33}^{r}}$)	~5000
Electromechanical coupling coefficient ($\boldsymbol{k_t}$)	0.60
Loss tangent	0.005

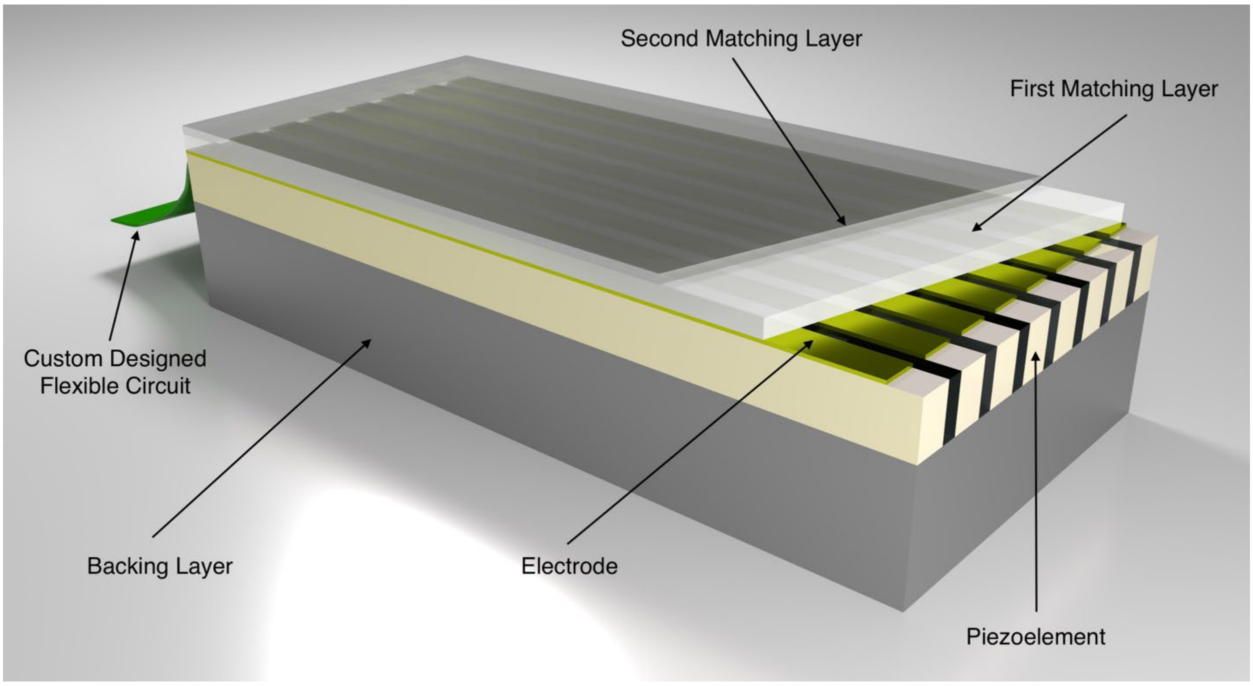


Figure 1. A schematic of designed phased-array transducer.

TABLE II. Properties of assembled layers

Layer	First matching layer	Second matching layer	Backing layer
Material	2-3µm Al_2O_3 powder + EPO-TEK 301-2	EPO-TEK 301-2	E-solder 3022
Weight ratio (Powder : Epoxy)	1.3:1	N/A	N/A
Acoustic impedance	7.6 MRayls	3.0 MRayls	5.9 MRayls
Longitudinal velocity	3320 ms^{-1}	2650 ms^{-1}	1850 ms^{-1}
Density	2295 kgm^{-3}	1132 kgm^{-3}	3200 kgm^{-3}

PiezoCAD (Sonic Concepts, Woodinville, WA) program based on KLM model was used to simulate and optimize the design of the transducer [28], [29]. The parameters of different layers were input to the simulation program, and then the optimized dimensions of the 20 MHz 64-element phased-array transducer were obtained as listed in Table III. Those dimensions were used to fabricate the array transducer in this study.

TABLE III. Optimized parameters of the phased-array transducer

Designed center frequency	20 MHz
Number of elements	64
Pitch	75 μm (1λ)
Elements width	55 μm
Elements length	5 mm
Elevation dimension	2.163 mm
Azimuthal dimension	4.8 mm
Kerf width	20 μm
PMNPT thickness	80 μm
First matching layer thickness	40 μm
Second matching layer thickness	33 μm
Backing layer thickness	2 mm

A PMN-PT single crystal was firstly coated with chromium (Cr) and gold (Au) electrodes whose thicknesses are around 200 nm and 500 nm, respectively, on both sides of the crystal plate by means of magnetron sputtering deposition system. The crystal plate was then poled under a DC field of 1 MV/m in silicone oil at $120^{o}\mathrm{C}$ for 15 minutes. The poled sample was then short circuited for 24 hours for removing the surface charges. The sample was waxed to a glass carrier plate and lapped to a thickness of 0.1 mm and then cut into a square of 5 mm × 5 mm by a mechanical dicing saw (DAD321, DISCO). The backing layer casted on the sample was prepared by using conductive epoxy E-solder 3022 (Von Roll Isola, New Haven, USA). After casted the conductive epoxy, the sample was put into a centrifuge with a rotating speed of 3000 rpm for 10 minutes. The conductive epoxy was then cured in oven for 24 hours at room temperature. The stack was then diced into a 32-element array with a dicing depth of $120\ \mu$m and a dicing spacing of $150\ \mu$m using the dicing saw. The dicing speed was 0.4 mm/s and the actual width of diced kerf is $27\ \mu$m. The kerf was subsequently filled with low viscosity epoxy EPO-TEK 301-2 (Epoxy Technology Inc., Billerica, MA) using capillary action for preventing elements collapsing and reducing the crosstalk between elements[30]. In some studies, kerfs were filled by a mixture of EPO-TEK 301-2 and 1-5 μm aluminum oxide particles to reduce the crosstalk significantly. [31] However, the large volume fraction of aluminum oxide particles would translate the filled kerf to a higher resonant frequency. So the high viscosity mixture is incapable to be used for narrow kerfs in this study. Then, the array was degassed in vacuum to remove air bubbles inside the kerfs. Lapping was processed after the filled epoxy was cured in an oven at room temperature for 24 hours. The array was diced again with the same spacing dimension by aligning the middle of elements as reference points, then the epoxy filling process was repeated. The 64-element array was lapped into the required thickness ($70\ \mu\mathrm{m}$). Alumina powder was used to polish the surface of the array for removing the scratches and enhancing the adhesion between piezoelements and electrodes. Then, the array was sputtered with Cr/Au (with a thickness of $200\ nm/500\ nm$) electrodes. By separating the electrodes, photolithography (Aligner 800MBA) was used to create an electrode pattern on the array.

The first matching layer, a mixture of 2-3 μm alumina powder and EPO-TEK 301-2, was prepared by centrifuging with a rotation speed of 3000 rpm for 15 minutes. After curing the matching layer with the previous method, the matching layer was lapped to the required thickness (40 μm). The second matching layer, EPO-TEK 301-2, was then casted on the lapped matching layer and degassed in vacuum to remove the air bubbles. The second matching layer was consequently lapped to a thickness of 33 μm. A designated flexible circuit with 64 electrode traces was aligned on the array under an optical microscope. A low viscous epoxy M-bond 601 was wiped on connecting points between the array and the flexible circuit, and the whole stack was clamped under a home-made metal fixture. M-bond 601 has higher mechanical stiffness than EPO-TEK 301-2 although both epoxies have similar viscosity, thus the flexible circuit can suffer higher stress from signal disconnection during utilization. The double matching layer was clamped on the surface of array by using the previous assembly method of flexible circuit. A coaxial cable was connected with the backing layer to serve as an anode. Eventually, a thin layer of parylene with a thickness of 3 μm was coated on the array transducer by using a parylene deposition system (Specialty Coating Systems, Inc., USA) for preventing water invasion and chemical hazards.

Results and discussions

Simulated Result

The simulated result of the designed array transducer is shown in Figure 2, where the pulse-echo response of the designed transducer presents a peak-to-peak value of 11.3 mV/V with a pulse length of 320 ns. It is apparent that the echo response is strong particularly for the small aperture size in phased-array transducer design. Besides, a very wide -6 dB bandwidth of 78% is shown from the echo response in the frequency domain. A center frequency of 21.7 MHz was calculated from the frequency spectrum.

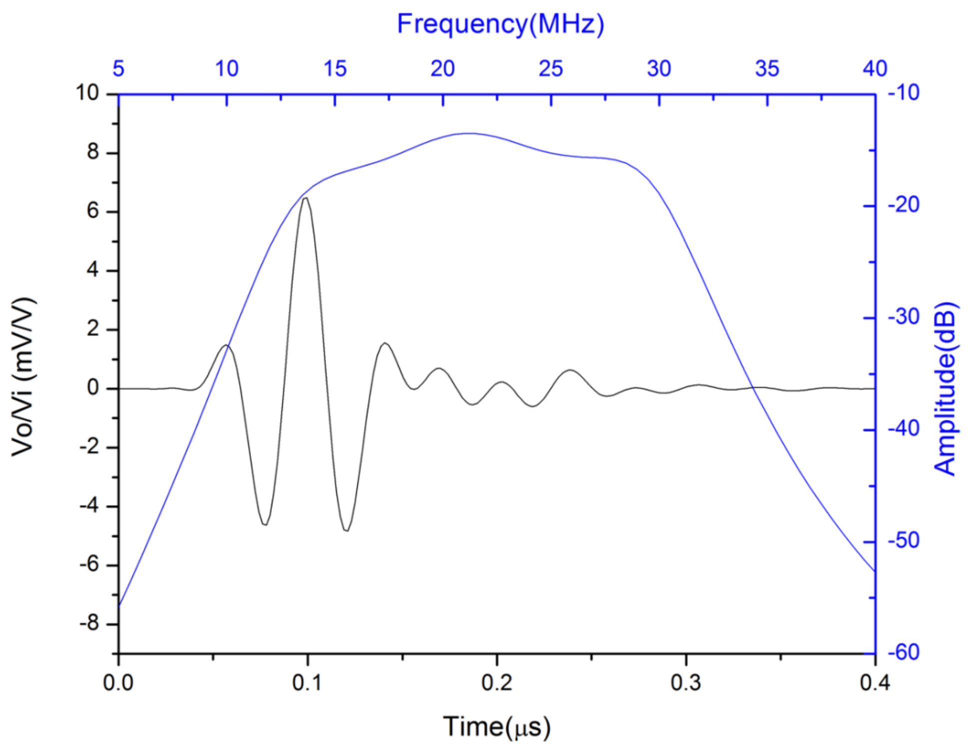
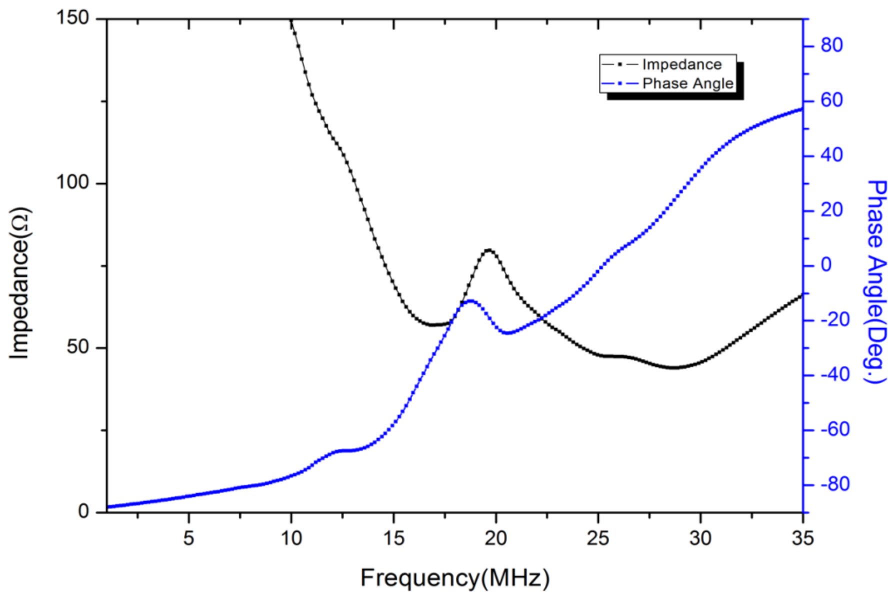


Figure 2. The simulated result of pulse-echo response (Black Line) and frequency spectrum (Blue line) by PiezoCAD.

Electrical Testing

Electrical impedance and phase angle response of the representative element of the phased-array transducer were measured in water by using an impedance analyzer (4294A, Hewlett-Packard). A 16047D test fixture (Hewlett-Packard) was used as the terminal of the analyzer, which was calibrated from 40 Hz to 110 MHz. The measured results shown in Figure 3(a) are obtained from a whole package including the flexible circuit connected with a PCB test board, and the coaxial cable connected with the backing layer. According to the phase angle response of the measured result, a strong phase transition peak is found at 18.6 MHz while two other weak transition peaks are found at 12.2 MHz and 26 MHz, respectively. Compared to the simulated result (transition peaks at 14, 22 and 29 MHz), which is illustrated in Figure 3(b), the measured peaks shift to lower frequencies with lower amplitudes. This may be attributed by the thickness variation of different layers between the simulation and experiments. Besides, the relatively weak transition peaks are probably due to the extra flexible circuit path of the developed array that is not included in the simulation. The comparisons of simulated and measured electrical impedance results are listed in Table IV. One can see that the measured electrical impedance of the representative element is 77.9 Ω at 20 MHz which is much closer to the standard impedance requirement of 50 Ω when compared to the simulated result.



(a)

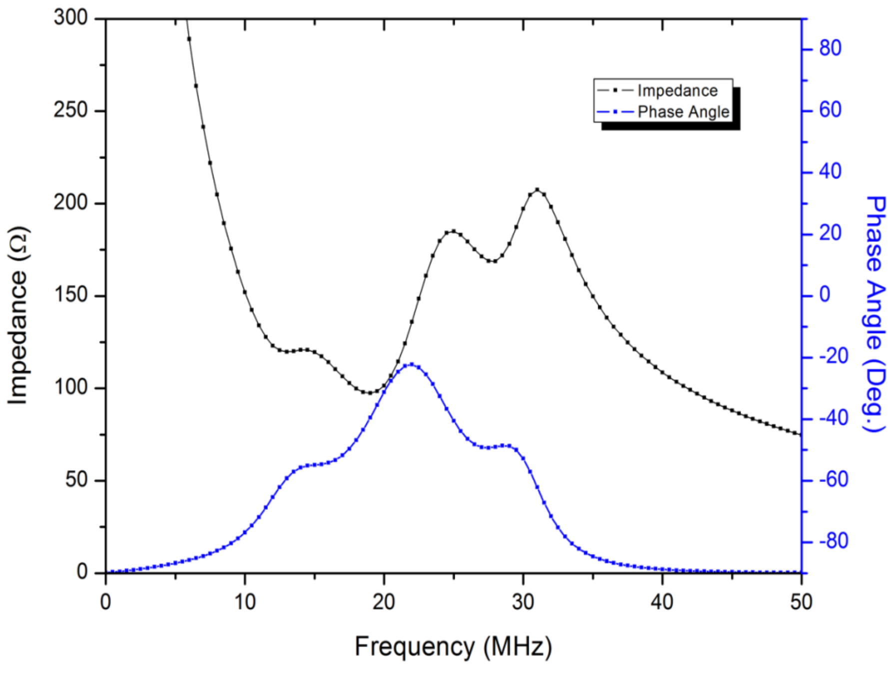


Figure 3.

(b)

TABLE IV. Comparison of simulated and measured results of electrical impedance

	PiezoCAD	Measured
Electrical impedance	101.0 Ω @20 MHz	77.9Ω @20 MHz
f_r	19.0 MHz	17.2 MHz
f_a	25.0 MHz	19.7 MHz
k_t	0.69	0.53

Acoustic Testing

The pulse-echo response of the representative element of the phased-array transducer was acquired in a water tank at room temperature using a Panametrics 5900PR pulser/receiver. The echo is reflected by a thick stainless steel target with a distance of 1 mm which is the near field-far field transition point. The active element is excited by an electrical impulse of 1 μJ at 500 Hz repetition with 50 Ω damping and 26 dB gain. The time domain echo response is displayed using an oscilloscope (HP Infinium), while the frequency domain echo response is calculated by the built-in Fast Fourier Transformation (FFT) program. For the insertion loss (IL) measurement, the amplitude of an echo signal V_0 reflected by a thick stainless steel target was measured by exciting the phased-array transducer connected with a function generator (AFG3251, Tektronix). The exciting waveform is a tone burst of 20-cycle sinusoidal wave with the amplitude of the driving signal (V_i) and the center frequency. An oscilloscope is connected with the excited phased-array transducer with 1 $M\Omega$ to measure V_0 and 50 Ω to measure V_i. The IL is calculated by using the following equation:

$$IL = 20\log(\frac{V_0}{V_i}) \qquad (1)$$

Figure 4 shows the measured pulse-echo response and frequency spectrum of the phased-array transducer, respectively. The echo exhibited a peak-to-peak value of 1.4 V with a pulse length of 314 ns. Besides, a -6 dB bandwidth of 91% was shown where the frequency ranges from 12 MHz to 32 MHz. The center frequency was calculated to be 22 MHz. The *IL* of the array was found as 29 dB. Table V shows the comparison between the measured and simulated results of pulse-echo and frequency spectrum. According to the comparison, the measured results were found to be slightly discrepancy with the simulated results. The measured center frequency makes very good agreement with the simulated frequency. The slight increment is due to the thickness discrepancy between the designed dimensions and fabricated dimensions as listed in Table VI. The experimental thickness of the piezoelement was 70 μm, so the measured center frequency was shifted to a higher frequency. For the echo amplitude, the measured result is higher than the simulated one. This should be mainly due to the impedance matching of the developed array transducer. With a thinner thickness, the designed piezoelement area is much more capable of achieving 50 Ω. Consequently, the developed array with a thinner and smaller piezoelement would give higher echo amplitude.

The most remarkable result of the phase array transducer is that the measured -6 dB bandwidth is significantly higher than the simulated result. It should be mainly attributed to the nature of piezoelement that is a 2-2 composite rather than a bulk single crystal. As known, the acoustic impedance of the 2-2 composite is lower so as to work even better with the existing double matching layer according to the theoretical calculation. Besides, when the volume fraction of the composite is larger than 0.1, the thickness mode electromechanical coupling coefficient of the 2-2 composites is higher than that of the bulk sample[32]. Consequently, the better impedance matching and higher coupling coefficient would offer wide-bandwidth in the measured results. The performance of the developed transducer shows significant enhancement when compared to the other reported 20 MHz array transducers fabricated using piezoelectric ceramic or ceramic-polymer composite materials[11], [24], [25].

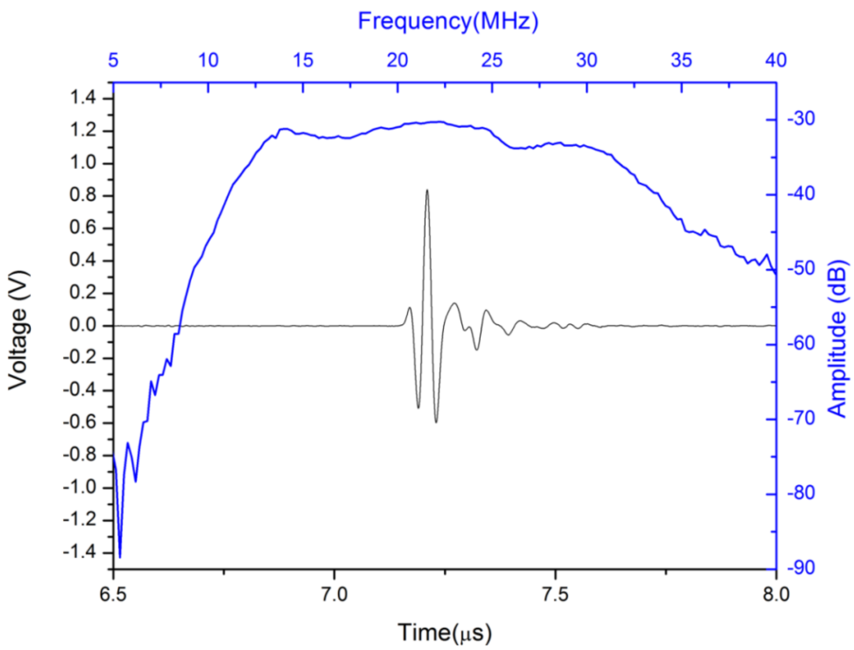


Figure 4. The measured result of pulse-echo response and frequency spectrum of phased array transducer.

TABLE V. Comparison of simulated and measured results of pulse-echo responce and frequency spectrum

	PiezoCAD	Measured
Number of elements	64	64
-6 dB Center frequency	21.7 MHz	22 MHz
-6 dB Bandwidth	78%	91%
Pulse length	320 ns	314 ns
Insertion loss	N/A	29 dB
V_{p-p}	11.3 mV (No Gain)	1.4 V (26 dB Gain)

TABLE VI. Comparison of parameters of designed and fabricated phased-array transducer

	Designed	Fabricated
Piezoelement thickness	80 μm	70 μm
Inner matching layer thickness	40 μm	38 μm
Outer matching layer thickness	33 μm	30 μm
Backing layer thickness	2 mm	2 mm
Kerf	20 μm	23 μm
Elements width	55 μm	52 μm

Conclusion

In this study, we have successfully demonstrated the development of a PMN-PT single crystal high-frequency phased-array transducer with kerfed 64-elements. The transducer exhibited a very wide -6 dB bandwidth of 91% and an insertion loss of 29 dB at its center frequency of 22 MHz. The small aperture size, and high bandwidth and sensitivity make the transducer very attractive for high-resolution imaging for the application of small animal studies. The kerf of the phased-array transducer is going to be reduced to further improve the performance of the transducer, and even higher frequency will be reached for the PMN-PT single crystal transducers when the laser micromachining technique is used.

Acknowledgements

We acknowledge the financial support by the Hong Kong Polytechnic University (Grant Nos. 4-ZZDC and G-YBLM) and the Hong Kong Research Grants Council (ECS Grant No. 253001/14P). This research was also supported by the National key Basic Research Program of China (973 Program) under Grant No. 2013CB632900 and the support from Shantou Institute of Ultrasonic Instrument (No: H-ZG4S).

References

[1] G. R. Lockwood, D. H. Turnball, D. A. Christopher, and F. S. Foster, "Beyond 30 MHz [applications of high-frequency ultrasound imaging]," *Engineering in Medicine and Biology Magazine, IEEE*, vol. 15, no. 6, pp. 60–71, 1996.

[2] C. Passmann and H. Ermert, "A 100-MHz Ultrasound Imaging System for Dermatologic and Ophthalmologic Diagnostics," *IEEE Trans. Ultrason., Ferroelect., Freq. Contr.*, vol. 43, no. 4, pp. 545–552, Jul. 1996.

[3] D. H. Turnbull, “In Utero Ultrasound Backscatter Microscopy of Early Stage Mouse Embryos,” *Computerized Medical Imaging and Graphics*, vol. 23, no. 1, pp. 25–31, 1999.

[4] F. S. Foster, J. Mehi, M. Lukacs, D. Hirson, C. White, C. Chaggares, and A. Needles, “A New 15-50 MHz Array-Based Micro-Ultrasound Scanner for Preclinical Imaging,” *Ultrasound in Medicine & Biology*, vol. 35, no. 10, pp. 1700–1708, Oct. 2009.

[5] M. Lukacs, J. Yin, G. Pang, R. Garcia, E. Cherin, R. Williams, J. Mehi, and F. Foster, “Performance and Characterization of New Micromachined High-Frequency Linear Arrays,” *IEEE Trans. Ultrason., Ferroelect., Freq. Contr.*, vol. 53, no. 10, pp. 1719–1729, Sep. 2006.

[6] H. D. Liang and M. J. K. Blomley, “The Role of Ultrasound in Molecular Imaging,” *The British Journal of Radiology*, vol. 76, no. 2003, pp. S140–S150, 2003.

[7] M. J. Zipparo and K. K. Shung, “Piezoceramics for High-Frequency (20 to 100 MHz) Single-Element Imaging Transducers,” *IEEE Trans. Ultrason., Ferroelect., Freq. Contr.*, vol. 44, no. 5, pp. 1038–1048, 1997.

[8] K. H. Lam, H. F. Ji, F. Zheng, W. Ren, Q. F. Zhou and K. K. Shung, “Development of Lead-free Single-element Ultrahigh Frequency (170 – 320 MHz) Ultrasonic Transducers,” *Ultrasonics*, vol. 53, pp.1033–1038, 2013.

[9] T. A. Ritter, T. R. Shrout, R. Tutwiler, and K. K. Shung, “A 30-MHz Piezo-Composite Ultrasound Array for Medical Imaging Applications,” *IEEE Trans. Ultrason., Ferroelect., Freq. Contr.*, vol. 49, no. 2, pp. 217–230.

[10] R. Chen, N. E. Cabrera-Munoz, K. H. Lam, H. S. Hsu, F. Zheng, Q. Zhou, and K. K. Shung, “PMN-PT Single-Crystal High-Frequency Kerfless Phased Array,” *IEEE Trans. Ultrason., Ferroelect., Freq. Contr.*, vol. 61, no. 6, pp. 1033–1041, May 2014.

[11] A. Nguyen-Dinh, L. Ratsimandresy, P. Mauchamp, R. Dufait, A. Flesch, and M. Lethiecq, “High Frequency Piezo-Composite Transducer Array Designed for Ultrasound Scanning Applications,” presented at the 1996 IEEE Ultrasonics Symposium. Proceedings, 1996, vol. 2, pp. 943–947.

[12] S. Michau, P. Mauchamp, and R. Dufait, “Piezocomposite 30MHz Linear Array for Medical Imaging: Design Challenges and Performances Evaluation of a 128 Elements Array,” *IEEE Ultrasonics Symposium, 2004*, vol. 2, pp. 898–901 Vol.2, 2004.

[13] M. Cannata, Q. F. Zhou, and K. K. Shung, “Development of a High Frequency (35 MHz) Linear Ultrasonic Array Using 2-2 Composite Elements [Biomedical Applications],” *IEEE 1992 Ultrasonics Symposium Proceedings*, vol. 2, pp. 894–897 Vol.2, 2004.

[14] C. E. Morton and G. R. Lockwood, “Evaluation of Kerfless Linear Arrays,” presented at the 2002 IEEE International Ultrasonics Symposium, 2002, vol. 2, pp. 1257–1260.

[15] J. M. Cannata, J. A. Williams, and K. K. Shung, “A Kerfless 30 MHz Linear Ultrasonic Array,” *Ultrasonics Symposium, 2005 IEEE*, vol. 1, pp. 109–112, 2005.

[16] B. Zhu, N. Y. Chan, J. Dai, K. K. Shung, S. Takeuchi, and Q. Zhou, “New Fabrication of High-Frequency (100-MHz) Ultrasound PZT Film Kerfless Linear Array,” *IEEE Trans. Ultrason., Ferroelect., Freq. Contr.*, vol. 60, no. 4, pp. 854–857, Apr. 2013.

[17] B. P. Zhu, D. W. Wu, Y. Zhang, J. Ou-Yang, S. Chen, and X. F. Yang, “Sol-Gel

Derived PMN-PT Thick Films for High Frequency Ultrasound Linear Array Applications," *Ceramics International*, vol. 39, no. 8, pp. 8709–8714, Dec. 2013.

[18] G. G. Harman, *Wire Bonding in Microelectronics; 3rd Ed.* New York, NY: McGraw-Hill, 2010.

[19] Q. F. Zhou, K. H. Lam, H. Zheng, W. Qiu and K. K. Shung, "Piezoelectric single crystal ultrasonic transducers for biomedical applications," *Prog. Mater. Sci.*, vol. 66, pp.87–111, 2014.

[20] K. H. Lam, Y. Chen, K. F. Cheung, and J. Y. Dai, "PMN–PT single crystal focusing transducer fabricated using a mechanical dimpling technique," *Ultrasonics*, vol. 52, no. 1, pp. 20–24, Jan. 2012.

[21] Q. Zhou, X. Xu, E. J. Gottlieb, L. Sun, J. M. Cannata, H. Ameri, M. S. Humayun, P. Han, and K. K. Shung, "PMN-PT Single Crystal, High-Frequency Ultrasonic Needle Transducers for Pulsed-Wave Doppler Application," *IEEE Trans. Ultrason., Ferroelect., Freq. Contr.*, vol. 54, no. 3, pp. 668–675, Mar. 2007.

[22] S. T. Lau, K. H. Lam, H. Chan, C. L. Choy, H. S. Luo, Q. R. Yin, and Z. W. Yin, "Ferroelectric Lead Magnesium Niobate-Lead Titanate Single Crystals for Ultrasonic Hydrophone Applications," *Materials Science and Engineering B-Solid State Materials for Advanced Technology*, vol. 111, no. 1, pp. 25–30, 2004.

[23] Y. Chen, K. H. Lam, D. Zhou, W. F. Cheng, J. Y. Dai, H. S. Luo, and H. L. W. Chan, "High Frequency PMN-PT Single Crystal Focusing Transducer Fabricated by a Mechanical Dimpling Technique," *Ultrasonics*, vol. 53, no. 2, pp. 345–349, Feb. 2013.

[24] E. Lacaze, S. Michau, and P. Mauchamp, "20 MHz Ultrasound Array for Medical Imaging From Design to Image Evaluation," presented at the 2001 IEEE Ultrasonics Symposium. Proceedings. An International Symposium, 2001, vol. 2, pp. 1139–1142.

[25] C. T. Chiu, J. A. Williams, B. J. Kang, T. Abraham, K. K. Shung, and H. H. Kim, "Fabrication and Characterization of a 20 MHz Microlinear Phased Array Transducer for Intervention Guidance," presented at the 2014 IEEE International Ultrasonics Symposium, 2014, pp. 2121–2124.

[26] S. Carey, C. Brox-Nilsen, H. Lewis, C. Gregory, and J. Hatfield, "Scanning Head with 128-Element 20-MHz PVDF Linear Array Transducer," *IEEE Trans. Ultrason., Ferroelect., Freq. Contr.*, vol. 56, no. 8, pp. 1769–1777, 2009.

[27] A. Macovski, *Medical Imaging Systems*. Prentice Hall, 1983.

[28] R. Krimholtz, D. A. Leedom, and G. L. Matthaei, "New Equivalent Circuits for Elementary Piezoelectric Transducers," *Electron. Lett.*, vol. 6, no. 13, p. 398, 1970.

[29] S. Sherrit, S. P. Leary, B. P. Dolgin, and Y. Bar-Cohen, "Comparison of the Mason and KLM Equivalent Circuits for Piezoelectric Resonators in the Thickness Mode," *ULTSYM-99*, vol. 2, pp. 921–926, 1999.

[30] A. R. Selfridge, "The Design and Fabrication of Ultrasonic Transducers and Transducer Arrays," Stanford University, 1982.

[31] J. M. Cannata, J. A. Williams, Q. F. Zhou, T. A. Ritter, and K. K. Shung, "Development of a 35-MHz Piezo-Composite Ultrasound Array for Medical Imaging," *IEEE Trans. Ultrason., Ferroelect., Freq. Contr.*, vol. 53, no. 1, pp. 224–236, Jan. 2006.

[32] W. K. Qi and W. W. Cao, "Finite Element Analysis and Experimental Studies on the Thickness Resonance of Piezocomposite Transducers," *Ultrasonic Imaging*, vol. 18, no. 1, pp. 1–9, Jan. 1996.