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Electroacoustic response of 1-3 piezocomposite transducers for high power applications

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Abstract

The electroacoustic performance of 1-3 piezoelectric composite transducers with low loss polymer filler was studied and compared to monolithic Pb(Zr,Ti)O 3 (PZT) piezoelectric transducers. The 1-3 composite transducers exhibited significantly high electromechanical coupling factor (k t ∼ 0.64) when compared to monolithic counterparts (k t ∼ 0.5), leading to the improved bandwidth and loop sensitivity, being on the order of 67% and -24.0 dB versus 44% and -24.8 dB, respectively. In addition, the acoustic output power and transmit efficiency (∼50%) were found to be comparable to the monolithic PZT transducers, demonstrating potential for broad bandwidth, high power ultrasonic transducer applications.

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Electroacoustic response of 1-3 piezocomposite transducers for high power applications

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The electroacoustic performance of 1-3 piezoelectric composite transducers with low loss polymer filler was studied and compared to monolithic Pb(Zr,Ti)O_3 (PZT) piezoelectric transducers. The 1-3 composite transducers exhibited significantly high electromechanical coupling factor ($k_t \sim 0.64$) when compared to monolithic counterparts ($k_t \sim 0.5$), leading to the improved bandwidth and loop sensitivity, being on the order of 67% and −24.0 dB versus 44% and −24.8 dB, respectively. In addition, the acoustic output power and transmit efficiency (∼50%) were found to be comparable to the monolithic PZT transducers, demonstrating potential for broad bandwidth, high power ultrasonic transducer applications. © *2012 American Institute of Physics*. [http://dx.doi.org/10.1063/1.4772482]

Piezoelectric transducers are widely employed in a range of medical applications, including ultrasound diagnostic imaging and ultrasound therapy. Ultrasound imaging is one of the most widely used imaging modalities due to the capability of a non-invasive visualization of interior tissues in real time with high resolution.[1] Recently, ultrasound has attracted attention for therapeutic applications, specifically ultrasound guided high intensity focused ultrasound (HIFU) therapy, as it allows for thermal ablation of malignant and benign tumors without open surgery.[2–5]

The main differences between ultrasound imaging and therapeutic applications are operating frequency and power level. Megahertz or above ultrasound is usually used for diagnostic imaging as the image resolution is proportional to operating frequency. The acoustic power and intensity levels are relatively low, being on the order of 0.05 W and 1–2 W/cm^2, respectively, in order to prevent the tissue damage. In contrast, the acoustic power and intensity levels of therapeutic ultrasound, such as HIFU, are several orders of magnitude greater, typically being on the order of 10–300 W and 1000 to 20,000 W/cm^2, with operating frequencies in the range of 0.8 MHz to 2 MHz. General characteristics of the ultrasound used for different therapeutic applications, such as frequency and acoustic intensity, are summarized in Fig. 1.[6]

For the generation of high power ultrasound, high electromechanical coupling k_{ij} and mechanical quality factor Q_m of piezoelectric transducers are desired, where the high electromechanical coupling factor permits effective energy conversion from electrical energy to mechanical energy, while the high mechanical quality factor reduces heat generation under high power operation. The electroacoustic power efficiency is also improved with increasing vibration velocity (v) of a piezoelectric material, which is related to the acoustic output power, and proportional to the product of electromechanical coupling and mechanical quality factor.[7]

Piezoelectric/polymer composites with 1-3 connectivity offer several advantages over monolithic ceramics for medical ultrasonic transducers, with high electromechanical coupling (k_{ij}), low spurious modes, and flexibility in terms of shaping.[8,9] However, the high power performance of 1-3 composites is generally lower than that of monolithic ceramics due to the inherently low mechanical quality factors and low thermal conductivity of the polymer fillers.[10–13] Approaches to improve the high power performance of composite transducers are, thus, to minimize the losses of the composite and/or to improve the thermal conductivity of the polymer matrix. Previous studies have shown that composites filled with a low loss polymer showed promising high power characteristics, exhibiting improved mechanical quality factor, consequently enhancing mechanical dynamic strain for a given drive field.[14] Composites comprised high thermal conductivity filled polymer, however, showed relatively low electromechanical coupling and mechanical quality factor, but exhibited comparable dynamic strain to high mechanical quality factor composites under high duty cycle conditions, possibly due to the improved heat management of the high thermal conductivity polymer.[14]

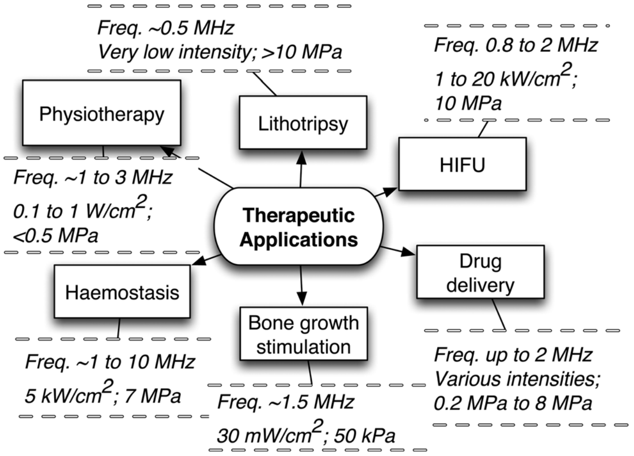

FIG. 1. General frequency, temporal-average intensity, and pressure amplitude of ultrasound used for different therapeutic applications (The data in this figure is from National Physical Laboratory Report DQL AC015, "Requirements for measurement standards in high intensity focused ultrasound (HIFU) fields," by A. Shaw and G. ter Haar, February, 2006.)

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The purpose of this work was to compare the electroacoustic performance of 1-3 composite transducers with those of transducers fabricated from monolithic piezoelectric ceramics under high power operation. Two different types of 1-3 composites were manufactured utilizing low loss polymer and high thermal conductivity polymer, respectively. Their high power performance was investigated under pulsed and continuous wave (*cw*) conditions in water.

Composites with 0.8–1.0 MHz fundamental resonant frequency were selected based on the typical operating frequency range of therapeutic transducers. In order to achieve high electromechanical coupling and mechanical quality factor, high volume percent (75%–80%) of piezoelectric material and low ratio of kerf width to post width were designed.[15,16] A commercially available "hard" lead zirconate titanate (PZT4, Piezokinetics, Bellefonte, PA) was chosen as piezoelectric active component, due to its high piezoelectric coefficient and mechanical quality factor. For the passive phase, two different types of polymers were used: (1) low elastic modulus and low loss polymer, (I) (Spurr resin, Polyscience, Inc.) and (2) high thermal conductivity polymer, (A) (T7110, Epoxy Technology, Inc.). Various properties of the two polymers, such as elastic modulus, thermal conductivity, viscosity, and attenuation, are summarized in Table I.[16] Poled PZT ceramics were diced using a Thermocarbon dicing saw (Tcar 864-1, Florida), with dicing parameters as follows: 0.100 mm ± 0.015 mm of kerf width, 0.800 mm ± 0.010 mm of post width, and 3.00 mm ± 0.025 mm of cut depth. The epoxy polymers were then backfilled into the kerfs and subsequently cured according to the manufacturers' instructions. The fabricated composites were polished until all the piezoelectric posts were exposed, and gold electrodes were sputtered on the large parallel surfaces. The dielectric $K_{33}{}^{T}$, piezoelectric d_{33}, acoustic impedance Z, mechanical quality factor Q_m, and electromechanical coupling k_t of the fabricated 1-3 composites are summarized in Table II. Note that PZT4 (I) composites showed relatively higher electromechanical coupling and mechanical quality factor compared to PZT4 (A) composites, owing to the relatively low elastic modulus and loss factor (related to the attenuation and viscosity) of polymer (I), as listed in Table I.

TABLE I. Elastic modulus (c_{ij}), viscosity (ν), thermal conductivity (k), and longitudinal attenuation (α_L) properties for various polymer materials. Elastic stiffness constants of the polymer materials were measured at 10 MHz for longitudinal and 5 MHz for shear.

Polymer	c_{11} (GPa)	c_{44} (GPa)	c_{12} (GPa)	ν (Poise)	α_L (dB/m)	k (W/mK)
(I)	5.28	1.08	3.12	0.65	2360	0.3
(A)	14.98	4.96	5.06	14–22	4100	1.0

TABLE II. Measured dielectric, piezoelectric, and electromechanical properties for PZT4 monolithic ceramics, PZT4 (I) and PZT4 (A) 1-3 composites.

Material	ρ (kg/m^3)	$K_{33}{}^{T}$	tan δ (%)	Z (MRayl)	d_{33} (pC/N)	Q_m	k_t
PZT4	7900	1540	0.56	36	350	400	0.50
PZT4 (I)	6270	1150	0.46	25	300	150	0.64
PZT4 (A)	6500	1130	0.42	27	290	50	0.58

For transducer design, the important considerations are acoustic and electrical impedance matching, which provide efficient power transfer between transducer/propagating medium and power source/transducer, respectively. Acoustic matching layers were prepared using the mixture of the polymer and 5 μm Al_2O_3 particles to get intermediate acoustic impedance values between water (~1.5 MRayl) and piezoelectric composites (~25–35 MRayl), with the thickness being down to the quarter wavelength according to

$$Z_m = \sqrt{Z_c Z_w}, \quad t_m = \frac{v_l}{4f}, \tag{1}$$

where subscripts c, m, and w denote the composite, matching layer, and acoustic media, respectively. f and v_l are the operating frequency and longitudinal wave velocity of polymers, respectively. In addition, the electrical impedance of the transducer was tuned and matched to the power electronics, which is usually on the order of 50 Ω. The doughnut shape magnetic core was used as an inductor to tune out the reactance of the transducers. The parameters of tuning inductor were calculated by using following equation:

$$L_p = \frac{|Z|}{\omega_c \sin[\theta]}, \quad n = \sqrt{\frac{R}{50}}, \tag{2}$$

where L_p is the parallel inductance, and ω_c is the center resonance frequency. Z and θ are impedance and phase angle at ω_c, respectively. n is the transformer turns ratio, and R is the resistance after tuning. The fabricated 1-3 composites and prototype transducers are shown in Fig. 2.

The performance of the prototype transducers was investigated using conventional pulse-echo response measurements, in which the fabricated transducers were placed in a water tank and excited by a high pulse voltage (−150 V) using a Panametrics pulser/receiver (model 5072PR, Panametrics Inc., Waltham, MA), with 200 Hz of pulse repetition frequency (PRF) and −35 dB amplifier gain. The pulse echo response of the transducers was recorded by receiving the reflected echo from the surface of a stainless steel target placed in the water bath. The reflected echoes were averaged and analyzed using a LeCroy WaveSurfer 452 oscilloscope. The loop sensitivity of the transducer was calculated

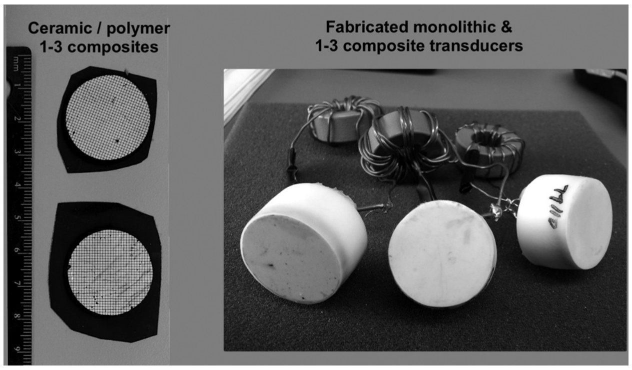

FIG. 2. Photographs of the ceramic/polymer 1-3 composites with 24 mm diameter, 0.8 mm postwidth, and 0.1 mm kerf width (left), and prototype transducers (right).

according to the following equation, which is defined as the ratio of received signal from the reflector to input voltage from the source:

$$\mathrm{SEN(dB)} = 20 \log \frac{V_r}{V_\mathrm{p}} - \mathrm{Gain}, \qquad (3)$$

where V_r is the peak-to-peak voltage of the received signal voltage and V_p is the pulse voltage (-150 V). For the measurements of transducer bandwidth, the fast Fourier transform (FFT) of the received waveform was stored, and the bandwidth was determined from the 6 dB down of the displayed amplitude spectrum according to the following equation:

$$\mathrm{BW}(\%) = 2\left(\frac{f_2 - f_1}{f_2 + f_1}\right) \times 100, \qquad (4)$$

where f_2 and f_1 are the higher and lower -6 dB frequencies.

For electroacoustic performance evaluation, the radiation force method was used,[17] as schematically shown in Fig. 3. A continuous sinusoidal wave was generated using HP 3314 A function generator and then amplified by a 50 dB power amplifier (MODEL ENI 325LA). A LeCroy WaveSurfer 452 oscilloscope was connected to the function generator in order to monitor the input voltage amplitude. The output power was measured using an ultrasound power meter (UPM-DT-10, Ohmic Instrument Co, Easton, MD), with measurement ability up to 30 W and a display resolution of ±2 mW. Note that the input power was derived from the measured voltage from a function generator, assuming that the electrical impedance (50 Ω) and phase angle (0°) of the transducers did not change with increasing drive voltage.

Fig. 4 shows the pulse-echo spectra of the prototype transducers. A transducer fabricated with monolithic PZT4 piezoelectric ceramics was also tested as a performance baseline. The most noticeable difference in pulse-echo signals between the PZT4 (I) composite and monolithic transducers was the achievable bandwidth, which was increased to about 67% from 44%, without sacrificing loop sensitivity level, being -24.0 dB versus -24.5 dB, respectively. Compared to the PZT monolithic transducers, improvements in both bandwidth and sensitivity of the composite transducers are deemed to be associated with their higher electromechanical coupling ($k_t = 0.64$) with low dielectric and mechanical losses, as evident in Table II. For the case of PZT4 (A) composite prototype transducer, the -6 dB bandwidth was found to be lower than that of PZT4 (I) counterpart, being on the order of 55%. The results suggest that the bandwidth of a transducer is directly proportional to the electromechanical coupling of a piezoelectric material, as PZT4 (A) composites have an intermediate value of coupling factor ($k_t \sim 0.58$) among the tested transducers. The loop sensitivity of PZT4 (A) transducers was found to be slightly lower than those of the monolithic and PZT4 (I) composite counterparts, being on the order of -26.3 dB, due to its relatively low mechanical quality factor. The results obtained from pulse-echo testing are summarized in Table III. It is evident that the piezoelectric composites with low loss polymer (I) can improve transducer performance in terms of bandwidth and sensitivity when compared to monolithic ceramic.

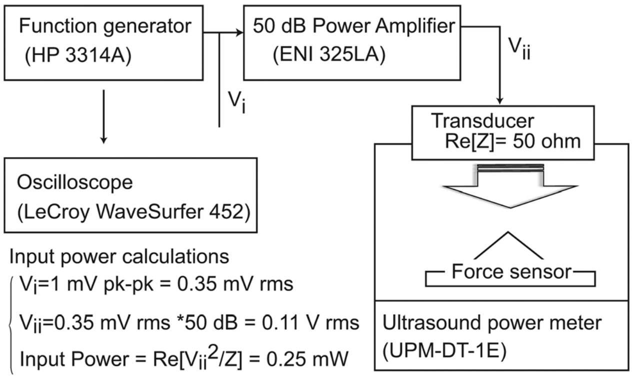

FIG. 3. Experimental setup for the measurements of acoustic input and output power of the transducers.

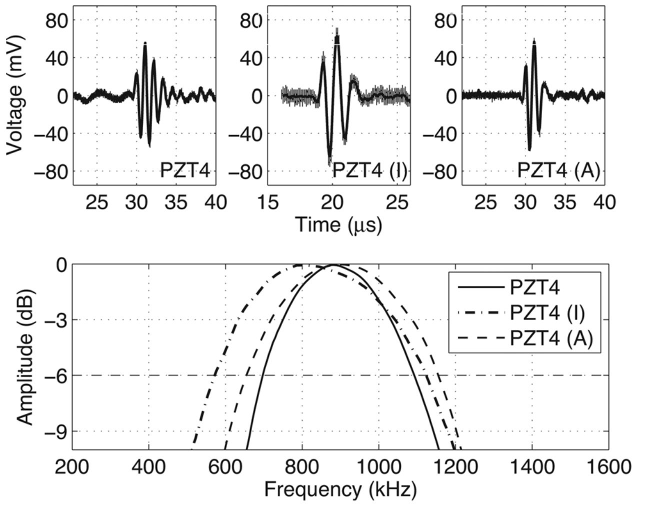

FIG. 4. Time-domain (up) and frequency-domain (down) pulse-echo responses from a steel reflector in water, using non-focused PZT4, PZT4 (I), and PZT4 (A) prototype transducers.

Fig. 5 shows the measured acoustic output power of the prototype transducers. It was observed that PZT4 (I) composites and PZT4 ceramic transducers showed similar acoustic output powers, reaching ~12 W at the input power level of 50 W. This reflects that the dielectric and mechanical losses of PZT4 (I) composite transducers are comparable to those of the PZT4 ceramic transducer. Interestingly, the composite transducer with high thermal conductivity polymer (A) showed relatively low acoustic output power, which gradually saturated with increasing input power, being on the level of about 4 W. The low level of acoustic output power of PZT4 (A) transducers is partly due to its relatively low electromechanical coupling and mechanical quality factor, since the acoustic power is proportional to the product of electromechanical coupling and mechanical quality factor.[7]

TABLE III. Measured transducer properties for PZT4, PZT4 (I), and PZT4 (A) prototypes. f_1 and f_2 are the higher and lower -6 dB frequencies. f_C is the center frequency.

Transducer	f_1 (kHz)	f_2 (kHz)	f_C (kHz)	Bandwidth (%)	Sensitivity (dB)
PZT4	698	1092	895	44	−24.8
PZT4 (I)	564	1126	845	67	−24.0
PZT4 (A)	652	1152	902	55	−26.3

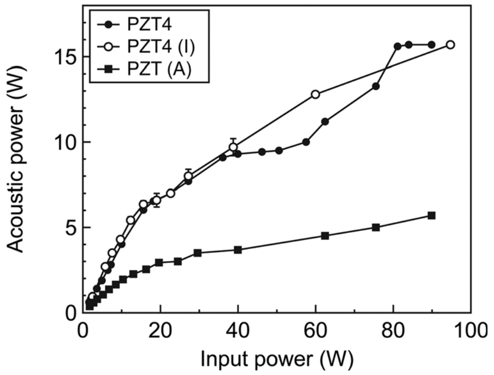

FIG. 5. Radiated acoustic output power of PZT4, PZT4 (I), and PZT4 (A) prototype transducers in water as a function of input power.

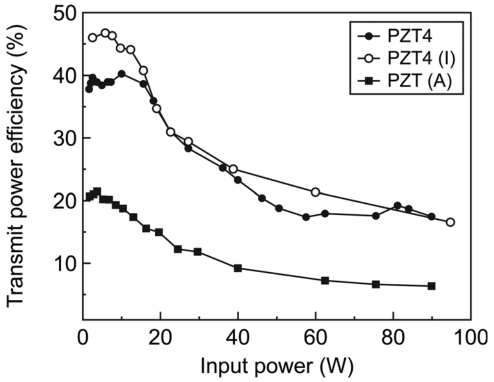

FIG. 6. Transmit power efficiency of PZT4, PZT4 (I), and PZT4 (A) prototype transducers as a function of input power.

Fig. 6 shows the transmit power efficiency of the various ultrasonic transducers. It was observed that the transmit power efficiency of all the PZT4 transducers decreases with increasing input power, which is related to the increased dielectric and mechanical losses under high drive conditions. Of particular significance is that the PZT4 (I) composite transducers showed the highest transmit power efficiency, offering ∼50% of initial transmit efficiency. In contrast, the transmit efficiency of PZT4 (A) prototypes was limited to ∼20%, which is believed to be associated with the higher mechanical loss of PZT4 (A) composites. This again, confirms that the high mechanical quality factor of the materials can improve the high power performance of the transducer in water-loading and cw conditions. However, it should be kept in mind that despite the reduced power efficiency, the use of high thermal conductivity polymers may allow more efficient heat flow into water under high power operation, improving long-term performance stability under high duty cycle (100%) condition. Thus, polymer fillers possessing high thermal conductivity, low elastic modulus, and low loss are desirable for high duty cycle, high power transducer applications, this study is underway.

In conclusion, PZT4 ceramic, PZT4 (I) composite, and PZT4 (A) composite prototype transducers were fabricated and compared. PZT4 (I) composite transducers were found to exhibit comparable loop sensitivity to PZT4 monolithic transducers, being on the order of −24 dB, while offering higher bandwidth, being on the order of 67%, compared to 44% for PZT4 ceramic prototypes. Together with its high radiated output power (up to 12 W at the input power of 50 W) and transmit power efficiency (∼50% initial transmit efficiency) under high drive condition, make the PZT4 (I) 1-3 composite potential for broad bandwidth and high power transducer applications, which will benefit both diagnostic imaging and ultrasound therapy.

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