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Lateral mode coupling to reduce the electrical impedance of small elements required for high power ultrasound therapy phased arrays

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Abstract

A method that uses lateral coupling to reduce the electrical impedance of small transducer elements in generating ultrasound waves was tested. Cylindrical, radially-polled transducer elements were driven at their length resonance frequency. Computer simulation and experimental studies showed that the electrical impedance of the transducer element could be controlled by the cylinder wall thickness, while the operation frequency was determined by the cylinder length. Acoustic intensity (averaged over the cylinder diameter) over 10 W/cm² (a therapeutically relevant intensity) was measured from these elements.

INTRODUCTION

Phased array technology has been utilized in diagnostic ultrasound for years. Phased arrays can dynamically focus and scan ultrasound beams without any mechanical movement, and they also correct for wave front distortions induced by the heterogeneous medium through which the wave is propagating. This static focus and distortion correction of ultrasound make high-powered arrays ideal for interventional purposes, such as thermal coagulation of tumors[1–5].

Three dimensional ultrasound field control requires a transducer element size to be small enough so that its emitted ultrasound field covers the entire volume in which the focusing is needed. Ideally, the element's center-to-center spacing should be equal to or less than half the wavelength, allowing for beam steering without grating lobes. The resulting small width-to-thickness ratio for such small elements results in a large electrical impedance and a mismatch between the RF-driving system and the elements. For high power arrays, this impedance mismatch has traditionally been addressed by adding an individual matching circuit for each transducer element [6]. This approach is reasonable for arrays up to a few hundred elements, but cost becomes a major factor when larger arrays are used. An ideal solution would be to manufacture a transducer element having an impedance close to the output impedance of the amplifier system, thus eliminating the need for additional electronics.

Electrical impedance mismatch has also been a problem with diagnostic phased arrays. One current solution involves assembling the transducer from multiple layers, allowing impedance reduction by increasing the electrode area and reducing the plate thickness [7].

However, manufacturing such transducers is fairly complicated, and their usefulness in high power applications has not been demonstrated.

In this paper, we explore a new method by which the impedance of high power phased array transducer elements can be reduced and controlled by a manufacturing technique [8]. With this method, the transducer elements are driven using the frequency of the length instead of the thickness mode. Transducer elements are manufactured from hollow, piezoelectric cylinders with electrodes on the inner and outer surfaces. Each cylinder is driven at the frequency that corresponds to its length mode resonance, thereby producing an ultrasound wave that propagates from the end of the cylinder. Our cylindrical design reduces the impedance of the transducer element by maximizing the surface area of the electrodes and reducing the thickness of the piezoelectric material between the electrodes. Since cylinder wall thickness can be modified without changing the length of the element, the electrical impedance of the element can be manipulated. We investigated the feasibility of our design using elements suitable for low frequency arrays designed for trans-skull sonications [9].

Material and Methods

Computer simulations

PZFlex (Weidlinger Associates, Inc, New York, NY, USA), a time domain finite-element-modeling (FEM) tool, was used to model the cylindrical transducer (fig. 1A). In figure 1, h is the length of the element, t is the wall thickness, and ϕ is the outside diameter. The boundary conditions of the simulation model were set such that the transducer was sitting on a backing material, immersed in water or air, with absorbing boundary around it. The backing material had an acoustic impedance of 1. MRayl and was highly damping (15dB/cm/MHz).

The transducer was made of PZT-4 ceramic with metal electrodes deposited on the inside and outside surfaces of the cylinder. The cylinder was polled in the radial direction such that the existing electrodes could be used in manufacturing. Radio frequency (RF) voltage at the length mode frequency (unless otherwise stated) as determined from the electrical impedance plots was applied across the thickness direction. The material properties used for simulation are shown in table 1.

Experiments

Transducers were constructed from radially polled, piezo-ceramic cylinders custom manufactured by EDO (Salt Lake City, UT, USA), and Valpey-Fisher (Hopkinton, MA, USA). Inner and outer surfaces of the cylinders were lined with metal (silver or nickel) electrodes. Dimensions of all of the cylinders are given in table 2.

Piezoelectric tubes were cut to the desired length using a diamond wire saw, then mounted on an acrylic plate, as shown in figure 1b. A layer of cork (thickness: approximately 2mm) was first glued to the plate, and then the cylinder was mounted on the cork with a layer of silicone rubber. This procedure reduces the wave propagation to the plate. This was verified by performing scanning laser vibrometer (PSV-400-M2-20 20MHz, Polytec, Tustin, CA, USA) measurements that demonstrated that the velocity amplitude of the wave propagated through the cork layer was only approximately 10% of the value at the end of the cylinder with the transducer element immersed in water. Copper wires were soldered to the outer and inner surface electrodes to provide electrical connections to a coaxial cable. The electrical impedance for all of the transducer elements and loading conditions were measured using a network analyzer (Model 4195A, Hewlett-Packard, Palo Alto, CA, USA). The electrical impedance plots were used to identify the different resonances of the cylinders. This was done by calculating the approximate resonant frequencies of each of the modes based on the

dimensions of the cylinder and then searching the impedance plots for the magnitude and phase peaks close to these frequencies.

The RF-signal feeding the transducer was produced by a frequency generator (Model DS345, Stanford Research Systems, Sunnyvale, CA, USA or, Model 395, Wavetek, San Diego, CA, USA) and an RF-amplifier (model 2400L, ENI Inc, Rochester, NY, USA). It was possible to generate power output from the transducers without any electrical matching. However, for the acoustic power and efficiency measurements, an external matching network was constructed and tuned for each transducer and frequency. This was done to assure accurate RF-power measurement. The forward and reflected RF-power was monitored by a digital power meter (Model 438A, Hewlett-Packard, Palo Alto, CA, USA) and a dual directional coupler (Model C1373, Werlatone, Brewster, NY, USA). The acoustic power output was measured using an in-house constructed, radiation force measurement system with an absorbing target [10]. Thickness mode measurements used a reflector to direct the cylindrical waves to the target [10], whereas for the length mode measurements the ultrasound waves, generated from the end of the transducer cylinder, were directed to the absorbing target. The 15 cm diameter target hung from an electronic balance (Model AE160, Mettler-Toledo, Inc., Columbus, OH, USA) with nylon lines. The balance was connected to a computer via an IEEE-488 interface. The computer captured weight measurements at 0.5 s intervals, before, during, and after a 15 s sonication, in order to calculate acoustic power (as described in [10]). Multiple measurements were performed at each power level before the applied RF-power was increased to the next level. All measurements were performed in deionized (providing electrical isolation in water immersion case) and degassed water (oxygen concentration < 1 ppm).

Ultrasound pressure wave distributions were measured using a needle hydrophone (spot diameter 1 mm) and an amplifier (Precision Acoustics Ltd, Dorchester, UK.). The hydrophone was moved by stepper motors in three dimensions under computer control. The hydrophone signals were captured by an oscilloscope (Model 2431L, Tektronix, Oregon, USA).

Results

Simulations

Simulation experiments investigated the electrical impedance and emitted ultrasound field as a function of the dimensions of the transducer element. A typical simulated phase and magnitude curve of the electrical impedance of a cylinder transducer (outside diameter: 10 mm; length: 10 mm; wall thickness: 1.2 mm) in air is shown in figure 3. There are three resonant modes: length mode, circular mode, and thickness mode. The thickness mode is a resonance in thickness direction because of the piezoelectric coupling of d_{33} . Circular mode is a resonance in circumference (radius) because of the piezoelectric coupling of d_{31} . Length mode is a resonance in the length (axial) direction, also because of the piezoelectric coupling of d_{31} . To avoid interference from other resonant modes, the dimensions of the transducer were chosen to avoid multiple vibration modes at frequencies too close to the desired length mode resonance. Length, circular, and thickness modes are marked as resonance peaks #2, #1, and #6, respectively, in figure 3. In addition, the third (#3), fifth (#4), and seventh (#5) harmonic resonant frequencies of the length mode are clearly visible in the graphs.

We tested the effect of cylinder wall thickness on the magnitude of electrical impedance at the length resonance frequency of these elements (fig. 4). Simulation studies were performed on two different cylinders with lengths equal to their outside diameter (10 mm and 20 mm, respectively). The results showed that the magnitude of the transducer

impedance increased with the increasing wall thickness. However, the length resonant frequency was only slightly dependent on the wall thickness.

We then simulated loading conditions of the cylinder in figure 3. Water loading decreased the electrical impedance magnitude peak, but had only a small impact on the phase (fig. 5A,C). Air loading, with the cylinder tip touching the water, produced magnitude and phase graphs that were in between the air-only and water-only curves. The maximum simulated electro-mechanical conversion efficiency for the 10 mm long and 10 mm diameter cylinder with 1.2 mm wall thickness was 34 %, at 182 kHz frequency.

Figure 6 shows the simulated length resonant frequency for 10 and 15 mm diameter cylindrical transducers as a function of the length of the cylinder. These cylinder dimensions were selected to match the experimental cylinder dimensions. The solid line is for the 10 mm diameter cylinder with the wall thickness of 1.2 mm. The dotted line is for the 15 mm diameter cylinder with 3.9 mm wall thickness. The graph shows that the length controls the frequency in a nonlinear fashion that is dependent on the cylinder diameter and wall thickness, as demonstrated by the slightly different shapes of the two curves (fig. 6). The symbols are for the experimental measurements with cylinders with the same dimensions as the simulation curves (see experiments).

Experiments

The measured phase and magnitude of the electrical impedance of two cylinder transducers (outside diameter: 10 mm; wall thickness: 1.2 mm; and lengths: 10 and 5 mm) in air are shown in figure 7. The measured curves demonstrate the length resonance frequency dependence on the length of the cylinder. The curves are similar to the simulated curves in figure 3.

Several different loading conditions (fig. 2) were tested to simulate conditions that would occur when therapy arrays are manufactured. The conditions tested were: the cylinder completely in air, water, epoxy, silicone rubber, or air with the end coupled to water. The latter test condition requires further explanation. In many ultrasound field measurements, a thin PVC membrane (thickness 0.07mm) is attached at the end of the cylinder to provide air loading in and around the cylinder while the transducer is immersed in water. This configuration simulates a situation where the array is constructed such that air is surrounding cylinders with only their ends coupled to water via a thin membrane.

The measured phase and magnitude of the electrical impedance for the length resonance at the fundamental frequency of the 10 mm long cylinder in different loading conditions is presented in figures 5 B and D. Again, the experimental curves agree well with the simulated curves (figure 5 A and C).

We next measured resonant frequency for 10 mm (closed circles) and 15 mm (open circles) diameter cylindrical transducers as a function of the length of the cylinder. The results (fig. 6) demonstrate how the length resonance frequency depends on the length of the cylinder.

The cylinders emitted ultrasonic waves from their ends when they were driven at or near their length resonance frequencies. We made ultrasound power measurements presented here using an external matching network. The matching networks were used to obtain reliable RF-power monitoring; the monitoring equipment operated on a 50 ohm power line. Thus, when the same transducer was used to measure the acoustic power, even at a slightly different frequency, the matching circuit was retuned to 50+/-1 ohm and 0+/- 1 Deg phase. Similar high acoustic powers were measured in experiments without the matching networks. When the acoustic power measurements were repeated at different frequencies, the

maximum efficiency (27%) was located approximately at the phase peak for both the completely immersed cylinder and for the cylinder that was in air with its tip touching water (fig. 8). When the tip was immersed, the efficiency was relatively constant between the phase peak and the magnitude peak, whereas when the entire cylinder was immersed in water, the efficiency was reduced at the magnitude peak, and the acoustic output was reduced by approximately 30% from its peak value at the magnitude minimum. The average maximum efficiency of seven similar transducers was $36 \pm 7\%$. In a separate experiment, the acoustic power from a 6 mm long cylinder was measured at the thickness (1.585 MHz) and length mode resonance (0.3135 MHz) and the conversion efficiencies were found to be $60 \pm 2\%$ and $42 \pm 2\%$, respectively.

We then measured the acoustic power output from the end of a cylinder transducer as a function of the driving RF-power (fig. 9). An approximately linear relationship was observed between the RF- and acoustic powers for both the fundamental length resonance frequency and its third harmonic (fig. 9A). When two water-filled transducers (length = 5 mm, wall thickness = 1.2 and 3.9 mm, diameter 10 and 15 mm, and driving frequency 0.323 and 0.285 MHz, respectively) were driven until they failed (fig. 9B), the maximum power measurements of 10.7 W and 11.3 W were recorded. The transducers failed due to electrode damaged at the locations where the electrodes were connected to the RF-wires. Finally, the use of silicone or epoxy (not shown) as a filling material reduced the efficiency at higher powers tested as is shown in figure 9C.

We measured ultrasound pressure amplitude distributions (no matching networks) (fig. 10A and 10B) for a cylinder transducer that was completely immersed in water; driving frequencies were at the electrical impedance magnitude minimum and maximum. The fields showed some variation as the frequency increased, but the ultrasound main beam shape remained approximately the same. The field was slightly more directed when a PVC membrane was used to provide an air surface, both inside and outside of the cylinder (fig. 10C). Similar results were observed with the other transducers tested. The ultrasound field directivity decreased with decreasing cylinder diameter and approached that of a point source at the diameter of a half-wavelength or smaller. This was demonstrated by the ultrasound pressure amplitude distribution measured for a 3 mm diameter cylinder with the length of 10.4 mm driven at its length mode at the frequency of 0.144 MHz shown in figure 11.

Discussion

We used simulations and experiments to demonstrate the use of lateral mode coupling to produce therapeutically useful ultrasound power outputs from cylindrical ultrasound transducers at their length resonance frequency, when the RF-electrodes were arranged for wall thickness mode operation. Power measurements translated to a maximum transducer surface intensity of 13.6 W/cm^2 (The maximum measured acoustic power averaged over a circular area with the outside diameter of the cylinder since this is the area occupied by the cylinder on an array surface), higher than what is currently used for most therapy applications [11]. Our approach allows for control of the transducer element impedance so that the desired impedance value can be achieved using small elements required for phased arrays with full wave control. This design can eliminate the need for expensive electrical impedance matching.

Resonant frequency was somewhat affected by cylinder size and wall thickness, as was shown by both simulations and experiments. As expected, the impedance was proportional to the transducer wall thickness and the frequency was inversely proportional to the length of the cylinder. These parameters allow achievement of the desired impedance for a given

element size by choosing an appropriate wall thickness. In addition, the cylinders were able to produce adequate power between the magnitude minimum and maximum, thus allowing additional freedom in fine tuning the impedance of the device for continuous wave applications.

Using the length mode resulted in a lower electro-acoustic conversion efficiency (approximately 30–40 %) than that observed with the higher thickness mode frequencies of the same cylinders (approximately 60%). However, the length mode efficiency was approximately the same as that observed with the thickness mode excitation (efficiency=30–40%) with an element size equal to or less than half the wavelength [12]. Therefore, the cost of achieving lower electrical impedance is that the length mode operation may reduce the electro-acoustic conversion efficiency when compare with thickness mode operation of large elements.

The motivation for this study was to develop practical array elements for a system that uses low frequency (100–600kHz) trans-cranial sonications for the disruption of the blood-brain barrier [9]. The current design requires a hemispherical array with a diameter of 25–30 cm. According to our simulations, adequate electronic steering would be achieved with the 10 mm diameter elements that were shown to work well in the length mode. We can produce elements smaller than half the wavelength that allow unlimited beam steering without grating lobes. Therefore, our new method is practical for its intended use. For example, an array could be constructed by using the tested transducer element design and mounting the transducer cylinders on a cork mat on an acrylic base. Then, a thin plastic membrane could be placed over the transducer array and fixed to the cylinder ends by epoxy. Securing the membrane at the edges of the array to the acrylic support structure would allow air to surround the cylinders, thus minimizing coupling between elements while allowing the ultrasound wave to propagate through the membrane to the coupling water in front of the array.

The therapeutic applications of our new transducer design are far-reaching. While this study explored the use of cylindrical transducers, the method should also be feasible with other transducer geometries. Although low frequencies were explored, this approach may also be useful at higher frequencies and thus may open up many new possibilities for both therapy and diagnostic array design and construction. The cylindrical elements provide some unique features for therapy array design, since the transducer material is only in the cylinder wall, leaving the center of the cylinder free for other uses. This could provide, for example, a location for imaging transducers or hydrophones. Diagnostic arrays might also be developed where the signal is transmitted by the cylinder and then received with a detector in the center of the cylinder. It may also allow concentric cylinders to be used to provide a multi-frequency array design. Another advantage of the cylindrical array element is the ability to use gas or fluid cooling to allow higher power outputs. This may be a significant advantage in applications where high power outputs are needed from small arrays, such as intra-cavitary applications [13].

In conclusion, this paper introduces a novel method of controlling the electrical impedance of ultrasound transducers and transducer elements by using cylindrical element design and length mode excitation. Both computer simulations and experiments verified the practicality of the method and demonstrated adequate acoustic power outputs for therapy applications.

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References

1. Ebbini ES, Cain CA. Multiple-focus ultrasound phased-array pattern synthesis: Optimal driving-signal distributions for hyperthermia. *IEEE Trans Ultrason Ferroelectr Freq Contr.* 1989; 36:540–548.
2. Wan H, VanBaren P, Ebbini ES, Cain CA. Ultrasound Surgery: Comparison of strategies using phased array systems. *IEEE Trans Ultrason Ferroelectr Freq Contr.* 1996; 43(6):1085–1098.
3. McGough RJ, Ebbini ES, Cain CA. Direct computation of ultrasound phased-array driving signals from a specified temperature distribution. *IEEE Trans Ultrason Ferroelectr Freq Contr.* 1992; 39(8): 825–835.
4. Chapelon JY, Faure P, Plantier M, Cathignol D, Souchon R, Gorry F, Gelet A. The feasibility of tissue ablation using high intensity electronically focused ultrasound. *IEEE Ultrasonics Symp.* 1993; 93CH3301–9:1211–1214.
5. Fjield T, Fan X, Hynynen K. A parametric study of the concentric-ring transducer design for MRI guided ultrasound surgery. *J Acoust Soc Am.* 1996; 100:1220–1230. [PubMed: 8759971]
6. Daum D, Buchanan MT, Fjield T, Hynynen K. Design and evaluation of a feedback based phased array system for ultrasound surgery. *IEEE Trans Ultrason Ferroelectr Freq Contr.* 1998; 45(2):431–438.
7. Goldberg RL, Smith SW. Optimization of signal-to-noise ratio for multilayer PZT transducers. *Ultrason Imaging.* 1995; 17(2):95–113. [PubMed: 7571210]
8. Hynynen K. A Method to Control the Electrical Impedance of Phased Array Elements. *Proc IEEE Ultrasonics Symp.* 2006:1052–1055.
9. Yin X, Hynynen K. A numerical study of transcranial focused ultrasound beam propagation at low frequency. *Phys Med Biol.* Apr; 2005 50(8):1821–1836. [PubMed: 15815098]
10. Hynynen K. Acoustic power calibrations of cylindrical intracavitary ultrasound hyperthermia applicators. *Med Phys.* 1993; 20:129–134. [PubMed: 8455491]
11. Hynynen K, Clement GT, McDannold N, Vykhodtseva N, King R, White PJ, Vitek S, Jolesz FA. 500-element ultrasound phased array system for noninvasive focal surgery of the brain: A preliminary rabbit study with ex vivo human skulls. *Magn Reson Med.* July; 2004 52(1):100–107. [PubMed: 15236372]
12. Hutchinson EB, Hynynen K. Intracavitary ultrasound phased arrays for prostate thermal therapies: MRI compatibility and in vivo testing. *Med Phys.* 1998; 25(12):2392–2399. [PubMed: 9874833]
13. Yin X, Epstein LM, Hynynen K. Noninvasive transesophageal cardiac thermal ablation using a 2-D focused, ultrasound phased array: a simulation study. *IEEE Trans Ultrason Ferroelectr Freq Control.* June; 2006 53(6):1138–1149. [PubMed: 16846146]
14. WWW.efunda.com/materials/piezo/material_data

Figure 1. Diagrams of the transducer design: (A) simulations, (B) experiments, and (C) a photograph of one of the transducers built.

Figure 2.

A diagram of the different experimental settings of the cylinder transducers.

Figure 3.

The simulated phase and the magnitude of the electrical impedance of a 10 mm outside diameter (OD), 1.2 mm wall thickness cylinder with the length of 10 mm in air as a function of frequency. The numbers show: #1: circular mode, #2: length mode fundamental, #3: length mode 3rd harmonic, #4: length mode 5th harmonic, #5: length mode 7th harmonic, #6: thickness mode.

Figure 4.

The simulated magnitude of the electrical impedance (top) and frequency (bottom) at the length resonances as a function of the wall thickness of the cylinders with the length and diameter of 10 and 15 mm.

Figure 5.

The experimental (A,C) and simulated (B,D) magnitude and phase of the electrical impedance of a 10 mm OD and 10 long cylinder with the wall thickness of 1.2 mm when surrounded by air, water or air with the cylinder end touching water as a function of frequency.

Figure 6.

The length resonant frequency of cylindrical transducers as a function of the length of the cylinder. The circles are the measured values for two different cylinder diameters (10 mm and 15 mm) with various lengths. The solid lines are simulated values for the same two cylinder diameters.

Figure 7.

The measured phase and the magnitude of the electrical impedance of a 10 mm outside diameter (OD), 1.2mm wall thickness cylinder with the length of 10 mm and 5 mm measured in air as a function of frequency.

Figure 8.

The magnitude (A), and phase (B) of the electrical impedance and acoustic output efficiency (C) as a function of frequency for a 10 mm OD and 10 long cylinder with the wall thickness of 1.2 mm when surrounded by water or air with the end of the cylinder touching water.

Figure 9.

The measured acoustic power output as a function of applied RF-power. A. a 10 mm OD and 10 long cylinder with the wall thickness of 1.2 mm at its length resonance and its third harmonic. B. a two 5 mm long cylinders with 1.2 mm and 3.9 mm wall thickness and OD of 10 mm and 15 mm, respectively. C. a 10 mm OD and 10 long cylinder with the wall thickness of 1.2mm immersed in water or silicone rubber.

Figure 10.

The measured 2D ultrasound pressure amplitude contour plots in water (normalized to the peak pressure amplitude). The fields were emitted from the end of a 10 mm OD and 5mm long cylinder with the wall thickness of 1.2 mm. A. The sonication at the electrical impedance magnitude maximum frequency with water immersion, B. at the magnitude minimum frequency with water immersion, and C. at the magnitude minimum frequency with air in and around the cylinders with membrane coupling to water.

Figure 11.

The 2D ultrasound pressure amplitude contour plots measured in water. The field was emitted from the end of a 10.4 mm long cylinder with the OD of 3 mm and wall thickness 0.3 mm immersed in water at the frequency of 144 kHz.

Table 1

Material properties used in simulation [14]

density (kg/m ³)	
ρ	7600
elastic ($\times 10^{-12}$ m ² /N)	
s_{11}	12.7
s_{12}	-3.86
s_{13}	-5.76
s_{33}	15.6
s_{44}	39.2
s_{66}	33
piezoelectric ($\times 10^{-12}$ C/N)	
d_{15}	511
d_{31}	-132
d_{33}	315
dielectric ($/\epsilon_0$)	
ϵ_{11}^T	1475
ϵ_{33}^T	1300

Table 2

The dimensions of all of the cylinders used in the studies.

Outside diameter mm	Inside diameter mm	Length mm	Material
3.0	2.4	10.4	PZT4
4.0	3.4	10.0	PZT5A
4.0	3.4	4.7	PZT5A
5.0	3.8	9.8	PZT5A
6.0	5.1	10.0	PZT5A
6.0	5.1	14.1	PZT5A
6.0	5.1	4.2	PZT5A
10.0	7.6	10.0	PZT4
10.0	7.6	5.8	PZT4
10.0	7.6	5.0	PZT4
10.0	7.6	4.6	PZT4
10.0	7.6	3.9	PZT4
10.0	7.6	2.9	PZT4
15.0	7.2	25.0	PZT4
15.0	7.2	12.5	PZT4
15.0	7.2	10.8	PZT4
15.0	7.2	5.0	PZT4
15.0	12.6	15.0	PZT4