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Design and Characterization of Dual-Curvature 1.5-Dimensional High-Intensity Focused Ultrasound Phased-Array Transducer

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Abstract

A dual-curvature focused ultrasound phased-array transducer with a symmetric control has been developed for noninvasive ablative treatment of tumors. The 1.5-D array was constructed in-house and the electro-acoustic conversion efficiency was measured to be approximately 65%. *In vitro* experiments demonstrated that the array uses 256 independent elements to achieve 2-D wide-range high-intensity electronic focusing.

I. Introduction

For extracorporeal high-intensity focused ultrasound (HIFU) array transducers, the aperture size is often required to be large so that sufficient acoustic energy can be generated and converge on the target after multiple media attenuation. Concentric-ring and sector-vortex phased-array applicators with a 12-cm aperture have been designed [1]. Larger heating patterns could be generated by the sector-vortex array while the concentric-ring array could be used for treating smaller volumes. Under some conditions, the concentric-ring array could produce a very large secondary focus along the central axis of the array. Combining the concentric-ring with sector-vortex configurations in the same transducer and using appropriate phases could suppress the effect of the secondary focus. The concept of a sparse, random phased array for focal surgery has been addressed [2]. Theoretical modeling showed that an array consisting of 64 circular, 8-mm-diameter, 2.1-MHz elements randomly mounted on a 100-mm-diameter spherical shell with 102 mm in radius of curvature could focus at 100 mm from the center of the array and could steer the focus to 2.5 mm and 5.0 mm off axis without the development of significant grating lobe levels. A 1.1 MHz, 256-element 12-cm-diameter spherical phased array with a 10 cm radius of curvature has been

developed for ablative treatment of large focal volumes of deep-seated tissue [3]. Magnetic resonance temperature images of an *in vivo* porcine thigh showed that a single hot spot could be shifted electronically along the depth direction by this phased array and a 5 cm³ tissue volume could be coagulated using a single HIFU exposure and five multiple-foci patterns switched at 18 Hz. A 1.0-MHz cylindrical phased array with an aperture of 15 × 12 cm has been designed and fabricated for the treatment of liver tumor [4]. 512 elements were used for a wide aperture, but only a 256-channel driver was used because the elements along the elevation direction were connected in parallel with respect to the center line of the array. The simulation results showed that the array had a dynamic focusing range from 145 to 175 mm and a steering range from –15 mm to 15 mm in the free field.

Our literature review showed a wide range of array qualities. The goal for the development of array-type HIFU transducers is to carry out precise and efficient ablation of tumors. The number of independent elements or driving channels must be as low as possible, considering the cost and complexity. The steering range of the ultrasound focus should be increased for various clinical applications. On the other hand, a focused phased array is preferred over a planar phased array for extracorporeal HIFU applications because it can generate higher intensity at the focus and lower intensity/damage on the acoustic window/skin [5]. Furthermore, the azimuthal steering range of a spherically focused phased array and the axial dynamic focusing range of a cylindrically focused phased array are limited, whereas the dual-curvature focused phased array can provide sufficient electronic-focusing ranges in azimuthal and axial directions. Therefore, a dual-curvature 1.5-D HIFU phased-array transducer was designed, fabricated, and tested in this study.

II. Materials and Methods

A. Structure and Principle

The geometric shape of the dual-curvature HIFU phased-array transducer is depicted schematically in Fig. 1. Resulting from the design of two curvatures, acoustic beams generated by the array can converge on two regions near two centers of curvature when the input electrical phase of all elements is the same. After the small tuning of all elements' phases, the acoustic beams converge on one region between two radii of curvature (R_{yz} and R_{xz}) and it can be moved in the z - or x -direction via further adjustment of phases. In this study, the focus represents the point of the peak intensity in the focal zone and the definition of the focal zone is the area of –6-dB peak intensity.

B. Design Consideration and Numerical Simulations

For the treatment of tumors deep inside body, the typical frequency of operation is 1.0 MHz. To reduce the channel/cost of the power amplifier, elements with symmetrical control can be used. In this study, two elements in the y -axis were connected to one channel. To avoid burning the skin or tissue in the near acoustic field during HIFU sonications, and to attain sufficient spatial average intensity (I_{SA}) at the focus, the transducer requires a wide aperture. However, grating lobes may occur as a result of the wider pitch obtained when increasing the aperture while using the same number of elements; therefore, the aperture size is restricted. The curvature ratio (CR), defined as the ratio of R_{yz} to R_{xz} , is optimized by considering the focus-steering range as well as the spatial average intensity at the focus.

In numerical simulations, the Rayleigh-Sommerfeld integral was used to obtain the ultrasound pressure field and then the intensity distribution. The numerical element size was equal to one quarter of a wavelength and the region of calculation extended from 10 to 30 cm in the axial direction and ±10 cm in the azimuthal direction. The phase of independent elements was calculated using the pseudoinverse method [5]. The density of the medium

was 1000 kg/m³ and the acoustic velocity was 1500 m/s. The operating frequency was 1 MHz and the surface acoustic intensity was assigned as 1 W/cm².

C. Fabrication Process of the Testing Sample and Array and Testing Equipment

A single-element, 1–3 piezocomposite sample with an aperture of 4 × 2.5 cm and two curvatures of 24 cm and 16 cm was manufactured at the laboratory for preliminary tests. The fabrication process of the testing sample and multi-element array included several primary steps: the dice-and-fill technology for the piezocomposite formation, lapping process to obtain the desired thickness, shaping process, electrode formation, cabling, and housing. A 50-impedance matching network was also developed using LC circuitry to allow more electrical power to be transmitted into each element. Impedance analyses of the sample and array were done with an impedance analyzer (65120B, Wayne Kerr Electronics, London, UK). A 256-channel driving system (custom-made by Advanced Surgical Systems Inc., San Diego, CA) was used to trigger all elements of the array. The acoustic power generated by the array was measured using a radiation force balance (RFB-2000, Onda Corp., Sunnyvale, CA) from which the electro-acoustic conversion efficiency of the array was obtained.

D. Phantom and In Vitro Pork Experiments

A reversible transparent silicone phantom (Sylgard 527 A&B silicone dielectric gel, Dow Corning Corp., Midland, MI) was used as an acoustic absorber to observe the focusing performance of the array. Pork tenderloin was chosen for the *in vitro* experiments because the material property was more uniform than other pork tissue. The silicone phantom or pork was immersed in degassed purified water and fixed to the mechanical holder of a 3-axis motion mechanism during HIFU sonications. The energy density E required to form a lesion in the pork can be calculated by

$$E = \frac{P_e \times \eta}{(A \times n)} \div I_s \times I_f \times t, \quad (1)$$

where P_e is the electrical power supplied to the transducer, η is the efficiency of the transducer, A is the area of an element, n is the number of the element, I_s is the surface acoustic intensity in the simulation, I_f is the peak intensity in the simulation, and t is the power supply duration.

III. Results and Discussion

A. Initial Design of the Dual-Curvature Array

The array had 512 elements where 64 independent elements were arranged along the x -direction and 8 elements in the y -direction were connected and grouped in pairs. The length (L_x), height (H_y), and R_{xz} of the array were 16, 10, and 24 cm, respectively, in the initial design. The results of the simulation showed that the array produced the maximum I_{SA} when the CR was equal to 50% [see Fig. 2(a)] but the corresponding z -axis dynamic focusing range was approximately 4 cm, as shown in Fig. 2(b). We decided to adopt a CR of 66.7%, i.e., $R_{yz} = 16$ cm, to achieve a z -axis focusing range of 6.6 cm with a compromise of a 10% decrease in peak I_{SA} . Figs. 3 and 4 demonstrate that the array can move the focus in the z - and x -directions. However, it cannot steer the focus in the y -direction at all because another focal zone is formed as the desired one is steered slightly (see Fig. 5). The criteria to determine the dynamic focusing range are the half-maximum intensity and the existence of grating/side lobes. The intensities were 814, 2055, and 904 W/cm² at (0, 0, 12.3 cm), (0, 0, 15.2 cm), and (0, 0, 20 cm), respectively. Furthermore, there were no grating/side lobes in the simulated field. Therefore, the dynamic focusing range along the z -axis was determined to be from 13 to 19 cm. Fig. 4 illustrates only one case among many simulation results of

steering the focus along the x -axis. The intensities were 1956, 1615, 1165, and 766 W/cm² when the steered focus was at (1, 0, 15.5 cm), (2, 0, 15.5 cm), (3, 0, 15.5 cm), and (4, 0, 15.5 cm), respectively. It was found that there were no grating lobes in the simulated field and the peak intensity at 4 cm off the axis was less than the half-maximum intensity. Consequently, the array could steer the focus from -3 cm to 3 cm along the x -axis.

B. Prototype of the Dual-Curvature Array

The center frequency of the single-element sample was 0.955 MHz with the bandwidth of 0.210 MHz through the pulse-echo measurement and the electro-acoustic efficiency was 65%. A prototype of the HIFU phased-array probe is shown in Fig. 6. Fabrication errors caused a gap at the center of the transducer and epoxy was used to fill the gap. The aperture size of the prototype was 17×10 cm and R_{yz} and R_{xz} were measured to be 17 and 24 cm. The average resonant frequency of 256 independent elements was 0.934 MHz with the standard deviation of 0.008 MHz. A small standard deviation implies that the electrical characteristics of each channel were uniform. After the assembly and housing processes, the impedance and efficiency of the array are not significantly different from those of the dual-curvature, single-element, 1-3 piezocomposite sample. A few elements have higher or lower resonant frequencies resulting from unavoidable differences in piezocomposite thickness produced by the manual lapping process. The bandwidth of 0.2 MHz can maintain the performance of the array if the resonant frequency shifts because of the temperature rise of the array during high-power operation.

C. Silicone Phantom and In Vitro Experiments

Silicone phantom experiments showed that the array could produce individual lesion at around (0, 0, 14 cm), (0, 0, 17 cm), and (0, 0, 20 cm); the lesions were formed initially at 6.0, 1.5, and 2.0 s after HIFU sonications when the intensity at (0, 0, 14 cm), (0, 0, 17 cm), and (0, 0, 20 cm) was estimated to be 116, 463, and 347 W/cm², respectively. The lesions were initially generated at the expected positions and gradually expanded to the near field. We also observed all lesions were at the x - z plane of the array at $y = 0$. Phantom experiments also demonstrated that the array could steer the focus sequentially at (-3, 0, 19 cm), (-1.5, 0, 19 cm), (0, 0, 19 cm), (1.5, 0, 19 cm), and (3, 0, 19 cm). Under these circumstances, there is no appreciable lesion induced by grating lobes or unexpected focusing in the silicone when the focus is electrically moved in the x - and z -directions. Fig. 7 is the simulated intensity fields of the real transducer model. The maximum intensity occurred at around (0, 0, 17 cm) and the range of the half-maximum intensity was between 14 and 20 cm, which is consistent with the phantom experimental range. Therefore, the key reason for the difference between the initial field analyses and phantom experimental results is the geometric variation between the real transducer and the initial design.

Four pork fillets were individually used for various ablation experiments and the fillets were cut open to observe the lesions after HIFU ablations. Fig. 8(a) shows that the energy required to induce a lesion in the tenderloin is electrical power of 130 W for 10 s with the front side of the fillet at a distance of 14 cm. Based on the results of the simulation in Fig. 7(b), the energy density to form a lesion was calculated using (1), where $P_e = 130$ W, $\eta = 0.65$, $A = 0.25 \times 1.25$ cm, $n = 512$, $I_s = 1$ W/cm², $I_f = 1442$ W/cm², $t = 10$ s; the energy density was 7616 J/cm² at (0, 0, 14 cm). When the fillet was at a distance of 17 cm, no lesion was formed until the electrical energy reached 130 W for 10 s as shown in Fig. 8(b); the dimensions of lesion were $2.5 \times 2.5 \times 15$ mm. According to the data ($I_f = 2105$ W/cm²) shown in Fig. 7(c), the energy density to form a lesion was calculated to be 11 118 J/cm² at (0, 0, 17 cm). The electrical power of 200 W for 2 s was sufficient to ablate the pork at 20 cm and the dimensions of lesions were $1.5 \times 1.5 \times 5$ mm, $2.5 \times 2.5 \times 7.5$ mm, $3.5 \times 3.5 \times 7.5$ mm, $3.5 \times 3.5 \times 12.5$ mm, and $5.5 \times 5.5 \times 12.5$ mm, corresponding to sonication

durations of 2, 3, 4, 5, and 10 s, respectively [see Fig. 8(c)]. Using the data ($I_f = 1293 \text{ W/cm}^2$) in Fig. 7(e), the minimum energy density to form a lesion was calculated to be 2101 J/cm^2 at (0, 0, 20 cm). The array steered the focus along the x -axis in the fourth fillet and, as expected, five lesions were formed, as shown in Fig. 8(d). The 2-D electronic focusing ranges of the array in the pork experiments are the same as those in the silicone phantom experiments. However, more energy is required to form a lesion in the pork because the attenuation of pork (180 dB/m at 1 MHz [6]) is higher than that of silicone (52.5 dB/m at 1 MHz , measured in this study). Moreover, we estimated that the threshold of energy density to form a lesion in the tenderloin pork is 2101 J/cm^2 . Higher energy density was needed to form a lesion at (0, 0, 14 cm) or (0, 0, 17 cm) in the pork because there were blood vessels and fat in front of the focus in these two cases. The precise threshold of lesion formation can be determined if the temperature of tissue can be monitored in real time by MRI thermometry. Five lesions in Fig. 8(d) are not in the same plane because the fillet had a slight angle on the z -axis of the array and the soft pork tissue was prone to distortion which creates an illusion of lesions not on the same plane. Based on many pork experiments, we estimate the array can generate an energy density of $11\ 118 \text{ J/cm}^2$ at the focus without any appreciable temperature rise of the array itself or near-field burn.

Numerical results and a series of experiments revealed that the dual-curvature design can confine the focusing location/range of the beams and hence small phase difference between elements is required for focusing and steering in the restricted range. The other advantage of the dual-curvature configuration is only 256 independent elements are needed to achieve dynamic focusing and focus steering in two directions.

IV. Conclusion

We have successfully developed a prototype 1.5-D HIFU phased-array transducer with dual-curvature configuration to achieve 2-D electronic focusing in this research. Numerical simulations and a series of *in vitro* experiments have verified that the dual-curvature array possesses 6 cm dynamic focusing range in the z -direction and 6 cm focus-steering range in the x -direction. As a future work, the array will be integrated with an MRI system for MR-guided HIFU treatment of tumors.

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Fig. 1. Schematic diagram of the dual-curvature high-intensity focused ultrasound phased-array transducer.

Fig. 2. Results of numerical simulations: (a) curvature ratio (CR) versus normalized intensity and (b) the effect of the CR on the dynamic focusing range in the z -direction. The CR is the ratio of R_{yz} to R_{xz} .

Fig. 3.

Acoustic field analyses. By adjusting the phase of the elements, the array was electronically focused at (a) 12.3 cm, (b) 13.25 cm, (c) 15.2 cm, (d) 17.15 cm, (e) 19.05 cm, and (f) 20 cm in the z -direction from the center of the array, with peak intensities of (a) 814 W/cm², (b) 1328 W/cm², (c) 2055 W/cm², (d) 1792 W/cm², (e) 1169 W/cm², and (f) 904 W/cm². The maximum intensity occurred at 15.2 cm and the dynamic focusing range was from 13 to 19 cm because the peak intensities at 12 and 20 cm were less than the half-maximum intensity.

Fig. 4.

The array steered the focus from (a) (1, 0, 15.5 cm) through (b) (2, 0, 15.5 cm) and (c) (3, 0, 15.5 cm) to (d) (4, 0, 15.5 cm). Peak intensities were (a) 1956 W/cm², (b) 1615 W/cm², (c) 1165 W/cm², and (d) 766 W/cm². The steering range was determined to be from 3 cm to -3 cm in the azimuthal direction because the peak intensity at (4, 0, 15.5 cm) was less than the half-maximum intensity.

Fig. 5.

(a) The array focused at (0, 0, 15.5 cm). (b) The array steered the focus to (0, 0.25, 15.5 cm) but the other focal zone at (0, -0.25, 15.5 cm) occurred at the same time. Peak intensities were (a) 2086 W/cm² and (b) 601 W/cm².

Fig. 6. Picture of the dual-curvature high-intensity focused ultrasound phased-array probe consisting of 16 pieces of 1–3 piezocomposite, acrylic housing, and 256 coaxial and 4 ZIF connectors.

Fig. 7.

Acoustic field simulations of the real transducer model. The array electronically focused at (a) 13.25 cm, (b) 14 cm, (c) 17.15 cm, (d) 19.05 cm, (e) 20 cm, and (f) 20.9 cm in the z -direction from the center of the array with peak intensities of (a) 925 W/cm², (b) 1442 W/cm², (c) 2105 W/cm², (d) 1596 W/cm², (e) 1293 W/cm², and (f) 1022 W/cm². The maximum intensity occurred at (0, 0, 17 cm) and the range of the half-maximum intensity was from 14 to 20 cm.

Fig. 8.

Four fillets of pork with dimensions of approximately $8 \times 4 \times 4$ cm were used for *in vitro* experiments. (a) The front side of the first pork fillet was located 14 cm from the array and five lesions were induced at five positions by the same power with different durations. (b) The second pork fillet was located 17 cm from the array and five sonications were triggered: 130 W/2 s, 130 W/3 s, 130 W/5 s, 130 W/7 s, and 130 W/10 s. The pork was moved between each sonication and only 130 W/10 s could induce the lesion. (c) The third fillet was located 20 cm from the array and five sonications were performed at five regions. Five individual lesions were obviously apparent. (d) The fourth pork fillet was located 19 cm from the array and was fixed. The HIFU hot spot was electrically steered along the x-axis and five lesions were formed with 1.5-cm spacing, according to the sonication plan.