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NEW FOCUSED ULTRASOUND TRANSDUCER FOR FAT CELL DISRUPTION

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Being an emerging body-shaping technology of fat cell disruption, high-intensity focused ultrasound has been investigated intensively in recent years for its favorable natures such as painlessness, safety and noninvasion. One of the major problems for the technology, however, is the overheating of transducers. In this study, we modified the transducer design in order to solve the overheating problem. We simulated the performance of the modified design by finite element analysis and fabricated the newly designed transducer. By measuring the actual performance data, we proved that the new design can effectively reduce temperature rise while keeping the acoustic intensity field unaffected.

Keywords: Focused ultrasound transducer; fat disruption; acoustic field; overheating.

1. Introduction

In recent years, with the improvement of living standards, obesity has become one of the major threats to human health. Conventionally, the only surgical way to achieve effective weight loss and body shape is by removing local fat deposits through liposuction.[1] However, it is a painful process and sometimes causes damages on human body. The recovery also requires long time. People are thus seeking simple and noninvasive methods for this purpose. Such motivation prompts the investigation of the method using therapeutic ultrasonic transducers. This method is mainly making use of the acoustic focal spot to disrupt and lyse the fat cells in order to achieve effective fat loss.[2]

High intensity ultrasound was first applied in medicine in 1980 to break up kidney stones, called lithotripsy. Over the past 25 years this technology has been shown to be safe and effective, and thus expanded its scope of applications to liver cirrhosis, cancer, ophthalmology and other areas of treatments.[3,4] In recent years, the same technology has been applied to the destruction of fat cells. Figure 1 schematically shows the basic principle of the technology. A transducer emits pulsating ultrasound waves which are directed into the unwanted fat tissue. The dome shape transducer allows the ultrasound to converge on a small focal point. Within this focal point, the synergies of waves create a strong mechanical effect that breaks up the

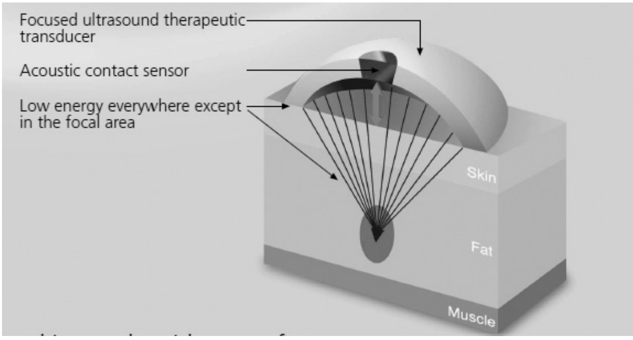


Fig. 1. Schematic drawing showing the basic principle of focused ultrasound therapeutic transducer.

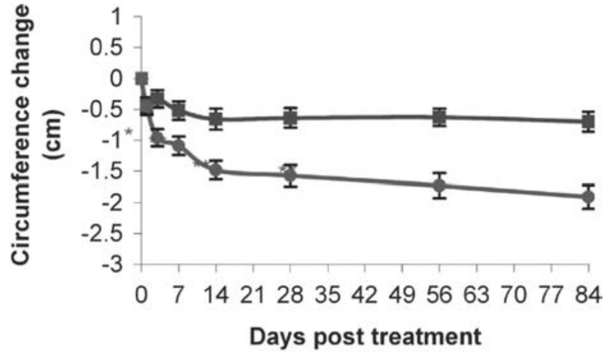


Fig. 2. Mean circumference change of the treated area versus mean circumference change of the internal control area (thigh).

fat cell membranes. As new fat cells are not produced in adulthood, fat cells that are destroyed will not reappear. This ensures that the treatment results are permanent. As opposed to liposuction, ultrasound only destroys the fat cells in the treatment area. Nerves and connective tissue are not affected. This is one of the reasons that the treatment does not cause any pain. One treatment lasts between 1 and 2 h depending on the size of the area and where it is situated in the body.

Clinical study showed that the focused ultrasound procedure reduced the circumference in the treated areas.[2] Average reduction in the circumference was approximately 2 cm in the abdomen, thighs and flanks. The reduction in circumference was corroborated by a reduction in fat thickness, as assessed by ultrasound measurement. The majority of the effect — 77% of the circumference reduction and 85% of the fat thickness reduction — was seen within the first 14 days after treatment, and additional reduction was seen over the following weeks. The effect was maintained for at least the study period of 12 weeks after a single treatment. One of the clinical data of mean circumference change of the treated area versus mean circumference change of the internal control area (thigh) is shown in Fig. 2. The circumference of the treated area was reduced significantly relative to the internal control area.

However, one of the major problems in the application is the overheating of ultrasonic transducers. Because the transducer works under high voltage for a long period during therapy, the temperature rise at the ceramic transducing element can be up to more than 100°C. The overheating brings about unfavorable experiences to the users, and even causes cracking failure of the ceramic transducer due to intensive thermal stress generated. In this study, we proposed a new transducer design to solve the overheating problem. The main idea was to increase the surface area of the ceramic element to

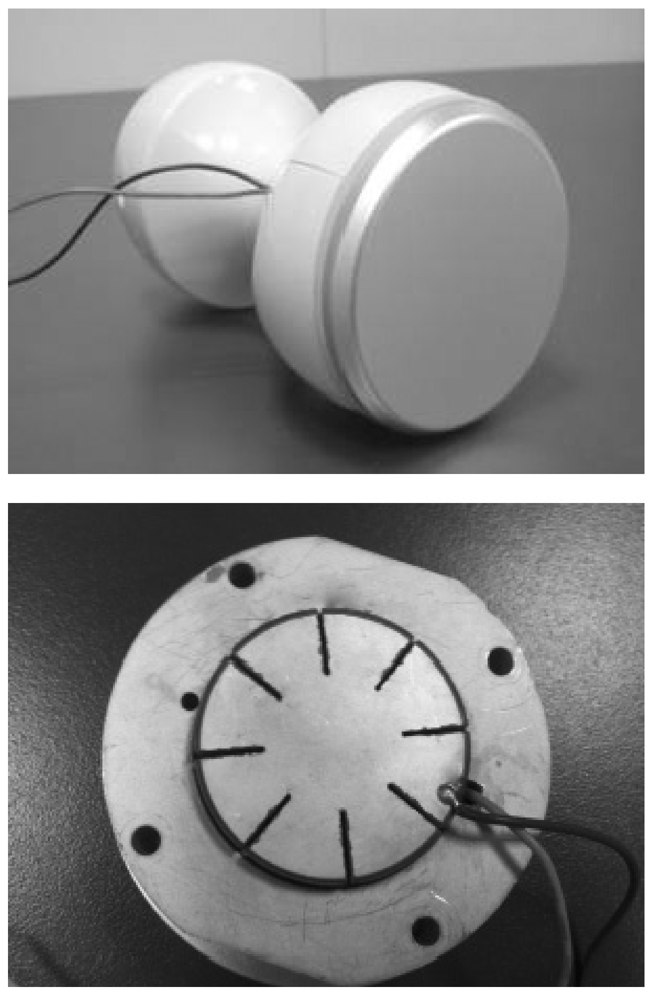


Fig. 3. The top view of the new structure transducer which had fabricated.

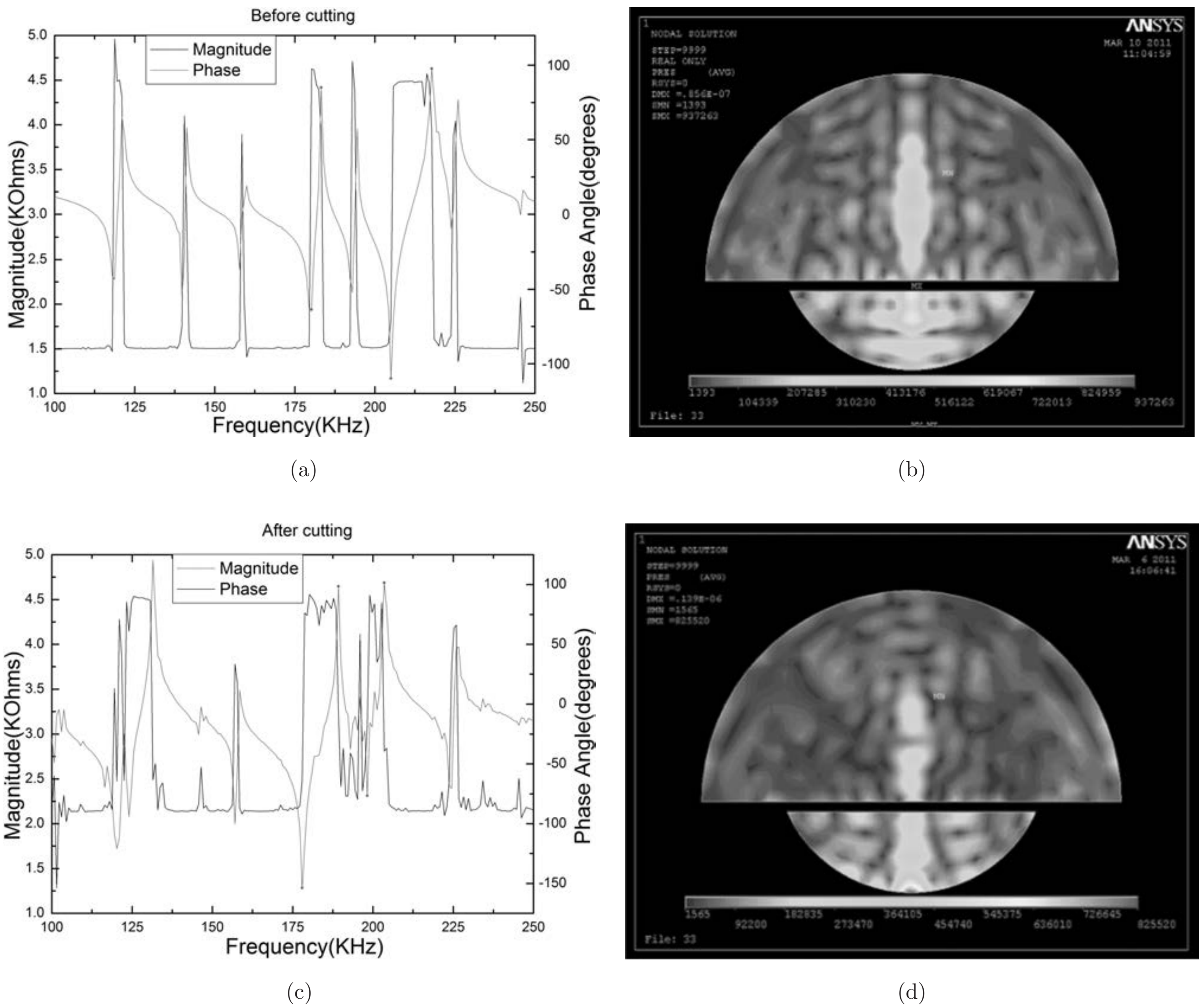


Fig. 4. (a) Impedance of ceramic wafer A and (b) packaged sound field distribution; (c) Impedance of ceramic wafer B and (d) packaged sound field distribution.

facilitate heat dissipation. The original Piezoelectric-42 ceramic plate having diameter of 44 mm and thickness of 11 mm was cut with eight slots to release thermal stress while keeping its acoustic field unaffected. Finite element analysis (FEA) was carried out to simulate the acoustic field of the new transducer. Then we measured the actual sound intensity field using radiation force balance (ONDA RFB-2000). The results showed that when the transducer was excited with AC voltage of 200 V peak at its resonant frequency, the max acoustical power of 12 W was achieved, and max acoustical strength was 31.84 W/cm^2. The temperature rise was also measured during high power excitation of the transducer. The results showed that the temperature was much lower than that of original transducer in the same working condition.

2. Experimental and Result

We modified the piezoelectric ceramic element by cutting the plate along eight symmetric directions, as shown in Fig. 3, in order to increase the surface area for heat dissipation as well as to effectively release thermal stress. The ceramic wafer was then attached to an aluminum acoustical lens epoxy 593. The front side of the acoustical lens was concave so that the ultrasound can be focused. Glycerol was filled in the concave as acoustic coupler. An aluminum cap was added to the top so that the glycerol will not get out. Figure 3 shows the top view of the modified transducer which was fabricated. Air backing was used here which was different from diagnostic transducers. Ceramic-to-air interface reflects most of the backward ultrasound wave to forward direction, and thus make sure there is less energy loss. Generally fat cells are located in about 5 ∼ 12 cm away from epidermis. It is expected that the focused sound field can be in this range. For simplicity in comparing the modified transducer with the original one, we use A standing for the original transducer, and B representing the modified one.

(1) Simulation

First, we checked the ceramic element's electromechanical property of transducer A using impedance analyzer (4294A, Agilent Technologies) in Fig. 4(a) and using FEA tool (Ansys) to simulate the transducer's ultrasound field (Fig. 4(b). Here we used the resonance frequency near 200 kHz. The semicircle above in Fig. 4(b) represents water which has no boundary, the below one represents the glycerol in the lens. The transducer works under the voltage of 200 V sine wave at resonant frequency. From simulation we found the place whose greatest intensity region is about 2−25 mm. Figures 4(c) and 4(d) show the ceramic piece B's electromechanical property and ultrasound field at the same excitation source. The intensity region in the water is about 0−20 mm. Although the depth of the ultrasound field is shorter than the original, the region of the center strength is the same. That is to say, the ability of disrupting the fat does not change.

(2) Acoustic field simulation testing

To verify the above simulation results, we measured the actual acoustic field of transducer B. First, we measured the sound field distribution in vertical direction. The transducer B was supported on the acoustic radiation force balance (RFB-2000 Tank, ONDA) using three-dimensional mobile platforms (KSA Series, Zolix) and excited with a 200 V sine wave at the resonant frequency. Reflective target was used here, moving the three-dimensional mobile platforms to the middle of the transducer surface, recording the field power at different depths, as shown in Fig. 5 (this result was the average of five measurements). Abscissa was the distance between the target and lens surface, the vertical axis measured sound intensity. The sound intensity peaks can be clearly observed at 0.5 mm, 5 mm and 9.5 mm which are basically consistent with the simulation results.

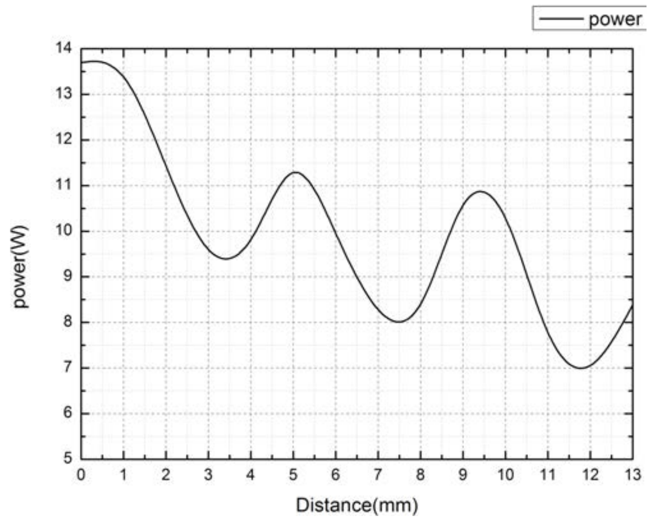


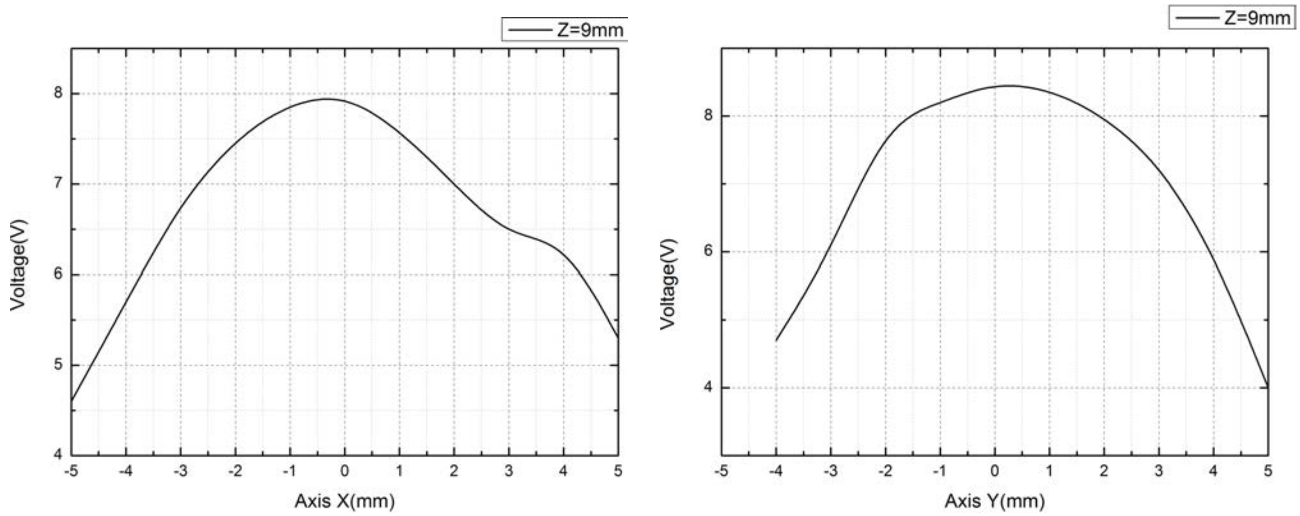
Fig. 5. Field distribution in the vertical direction.

The designed distance for main fat disruption of this transducer is about 10 mm away from epidermis. The modified transducer's sound field was enough to reach the targeted treatment area. We measured the focal spot size by using a micro ultrasound receiving probe. The receiving probe used a small pMUT transducer which has a resonant frequency of ∼200 kHz and size of 2 mm * 2 mm. Receiving signals were read out from the oscilloscope (TDS1012B-SC Tektronix). Sensor output voltages shown in the oscilloscope indicated the intensity distribution of the acoustic field. Figure 6 shows the voltage at the depth of 9 mm. Figures 6(a) and 6(b) represent the field distribution along two perpendicular directions. 0 mm represents the centerline.

If counted the focal spot within −3 dB, the focal spot area was about 37.68 mm^2 and the max intensity was calculated as 31.84 W/cm^2.

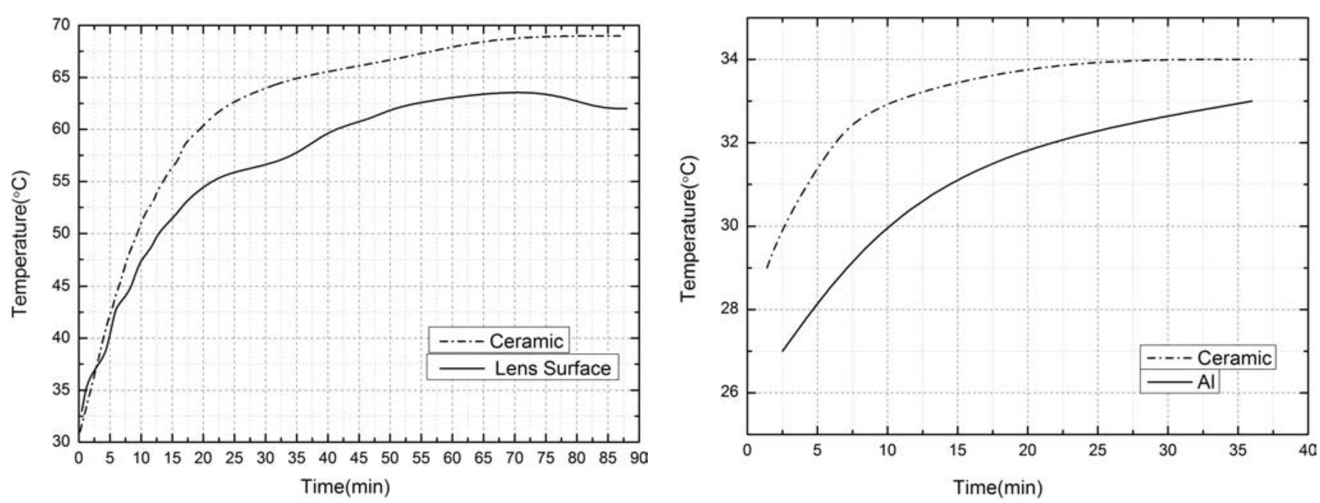
(3) Temperature rise test

Temperature rise of the transducer working under high voltage for long time was recorded in the test. Two thermocouples with same specification were attached to both the ceramic surface and the acoustic lens. Figure 7 respectively shows the testing results of transducer A's and B's



(a) (b)

Fig. 6. The sound field in two perpendicular directions at 9 mm from transducer surface.



(a) (b)

Fig. 7. The temperature rise against time for transducer A (a) and B (b).

temperature rises against time. It can be clearly seen that, after 20 min transducer B's temperature rise (6°C) is much smaller than transducer A's temperature rise (30°C). The temperature of the acoustic lens of transducer B is kept even lower than 35° for the entire testing period of 40 min, which means it is quite safe and will not bring about any unfavorable effects to the users.

3. Discussion and Conclusion

Although the acoustic field changed after the modification of ceramic element of the transducer, the sound intensity and the size of the focal spot were still meeting the target requirements. The resonant frequency of the modified transducer was 189 kHz. The focal spot is about 9 mm away from the transducer and the acoustical strength at the focal point is about 31.84 W/cm^2. Temperature rise was dramatically reduced by using the modified design. It was proven to be safe and effectively prevented from cracking failure.

Acknowledgments

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