Split-focused ultrasound transducer with multidirectional heating for breast tumor thermal surgery

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This study investigated the feasibility of using a split-focused ultrasound transducer to perform thermal surgery on breast tumors, based on a multidirectional heating scheme. The transducer is a square section of a sphere with a radius of 10 cm. The transducer was tilted such that its acoustic beam was 45° relative to the rib surface, and its focal zone was arranged by a shift of 6 mm away from the center of the planning target volume. The multifocus switching technique was employed to enhance the heating efficiency. When a single transducer was used, the transducer sonicated from a certain position for a given duration, and then rotated sequentially to continue the heating. Computer simulations and *in vitro* phantom experiments have been studied for this heating system. Both simulation and experimental results demonstrated that the system based on a multidirectional heating scheme is capable of generating a proper thermal lesion within 8 min. Meanwhile, from the simulation results, the rib heating was effectively alleviated by tilting the transducer to induce the total reflection at the muscle/bone interface. While using multiple ultrasound transducers, an appropriate arrangement was designed to have the same configuration of acoustic beams as is used for a single-transducer strategy. The simulation results from the four-transducer strategy indicated that the heating results could be further improved. This study revealed that it is very promising to have an appropriate arrangement of a single split-focused ultrasound transducer with mechanical rotation, or to have multiple split-focused transducers that use multidirectional heating for breast tumor thermal therapy. © 2008 American Association of Physicists in Medicine.

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I. INTRODUCTION

Focused ultrasound surgery (FUS), which can be a noninvasive approach for tumor ablation, has become more and more promising in recent years. This approach is able to deliver acoustic energy in a predetermined region and raise the temperature above the threshold of tissue coagulation or necrosis. Due to the clinical potential of localizing a thermal dose to a specific region through the intact skin, ultrasound thermal therapy has been evaluated in recent studies aimed at different sites such as liver tumors,¹ prostate cancer,^{2,3} breast cancer,⁴⁻⁶ kidney cancer,^{7,8} uterine fibroids,^{9,10} and bladder tumors.¹¹

Because of the absence of large blood vessels, breast tumors are suitable for applying ultrasound thermal surgery.

Additionally, the convex shape of the breast facilitates entrance of ultrasound beams through a large acoustic window. Specific designs with cylindrical ultrasound transducers¹²⁻¹⁵ for breast tumor treatment have been studied, and a highly focused spherical transducer has been used for breast tumor thermal surgery.⁵ The major difficulty encountered when implementing this approach is the small volume of the focal zone in comparison with the entire tumor size, thus requiring numerous sonications to accomplish the treatment. To enlarge the focal zone and enhance the efficiency of thermal surgery, a scheme using phased array to generate and switch a set of multifocus patterns was proposed in a previous study.¹⁶ In addition, clinical trials based on phased array to produce multiple foci for breast thermal surgery have also

been conducted.¹⁷ However, according to the clinical reports,^{6,18} it took an average of 1.3 h or longer to treat tumors with a mean size of 3.1 cm in diameter. In addition to the small size of the focal zone, the cooling interval between successive sonications was another factor accounting for the long treatment time. The cooling interval is needed to prevent the normal tissue in front of the tumor from overheating, and this procedure prolongs the treatment dramatically. In a previous clinical report,¹⁷ the cooling interval lasted up to 90 s to reduce the risk of skin burn.

Another major obstacle to applying FUS to breast tumors is the overheating of ribs, due to its high acoustic absorption. For tumors close to the ribs, the acoustic beam from a spherical ultrasound transducer will impinge upon the ribs during the heating and may induce a localized high temperature, resulting in patient discomfort. However, an acoustic beam will reflect at the bone/soft tissue interface due to the great difference in acoustic impedances, and an acoustic power deposition model accounting for the phenomenon of reflection and refraction at the muscle-rib interface was proposed.¹⁹ This model suggests that power reflection is a dominating factor for thermal dose delivery on the chest wall in ultrasound hyperthermia. Moreover, for the problem of rib overheating, a preliminary study²⁰ was proposed to optimize the thermal dose distribution in the breast and minimize the rib heating by using a hemispherical ultrasound phased array composed of 191 elements.

In order to effectively reduce the long treatment time and prevent the ribs from overheating, a split-focused ultrasound transducer with a multidirectional heating scheme was developed in this study. First, a four-element spherical ultrasound transducer was arranged with a tilt angle of 45° relative to the rib, and its focal zone was shifted away from the symmetrical line of the planning target volume (PTV). Second, three split-focus modes were temporally switched to form an enlarged thermal lesion. Finally, the transducer was rotated to sonicate the PTV from different directions. It was expected that a large tilt angle would allow an effective reflection to reduce the acoustic absorption of the ribs. Most importantly, it did not need a cooling interval between sonications, since the acoustic beams emitting from different paths minimized the overlap of the acoustic power in the normal tissue.

II. MATERIALS AND METHODS

II.A. Ultrasound transducer and heating unit

The ultrasound transducer used in this study is a square section of a sphere with a radius of 10 cm. The side length and driving frequency of the transducer are 7 cm and 0.5 MHz, respectively. The transducer was split into four equal elements, and three different modes of focal power patterns could be produced, as shown in Fig. 1(a), by changing their relative phases; the phases of elements for different modes²¹ are listed in Table I. A heating unit of $1 \times 1 \times 2$ cm³ was designed for each sonication, and to achieve dose optimization within the heating unit, a temporal switching and an optimal power weighting computation technique¹⁶

were employed to obtain a suitable set of power for the three focal patterns. While using this technique, the switching frequency and heating time were primarily set to 10 Hz and 10 s, respectively. The quasi-newton method²² was used to determine the optimized power intensity on the transducer surface for each pattern, as listed in Table I. The heating unit produced by the pattern switching is shown in Fig. 1(b).

II.B. Heating strategies

II.B.1. Single transducer with mechanical rotation

In this study, total reflection was used to alleviate the acoustic power deposition on the ribs.¹⁹ Considering the symmetry of thermal dose alignment and the total acoustic reflection at the soft tissue/bone interface, an angle of 45° between the central axis of the transducer and the normal direction of the tissue/bone interface was selected. Figure 2(a) shows the arrangement of the breast and transducer. The breast was modeled as a hemisphere, 12 cm in diameter. The ultrasound transducer was tilted 45° toward the breast, and water acted as a coupling medium in the space between the breast and transducer. To examine the extent of rib heating, the chest wall was modeled as a slab of bone with 1 cm thickness above the breast. The PTV inside the breast was modeled as a cube of $2 \times 2 \times 2$ cm³, and therefore, four heating units with dimensions of $1 \times 1 \times 2$ cm³ each were required to complete the treatment. To minimize the overheating of normal tissues in front of the PTV, the transducer was arranged such that sonication occurred from different directions to effectively use the acoustic window of the convex shape of the breast. Figure 2(b) shows the top view of the arrangement for sonication paths, which are centered at the PTV. The sonications are deployed separately around the breast to avoid overlapping the acoustic beam propagating through the normal tissue. For the purpose of distributing the thermal dose evenly, the focal center of the acoustic beam was 6 mm away from the geometrical center of the PTV. For different locations of tumors, the heating system was moved mechanically to make the rotation center coincident with the PTV's geometrical center.

In the simulation of a single transducer, the effect of conduction among heating units would have resulted in excessive dose accumulation if applied for four consecutive sonications. Therefore, we adjusted the heating duration from 10 to 8 s for each heating unit in order to achieve a proper thermal dose distribution for the entire PTV.

II.B.2. Four transducers with simultaneous sonication

A multidirectional heating scheme with four transducers was also investigated by simulation. We deployed four transducers to the positions shown in Fig. 2(b) to heat the PTV for 40 s simultaneously. The output power for each transducer was one-quarter of the original setting value (Table I) and thus the total energy delivered was equivalent to that for four heating units.

FIG. 1. A square focused ultrasound transducer with radius of curvature of 10 cm with driving frequency of 0.5 MHz and side length of 7.0 cm. The transducer is split into four equal square elements and driven with different modes to form different focal power patterns. (a) Three focal power patterns on the focal plane produced by the transducer. (b) A $1 \times 1 \times 2$ cm³ heating unit produced by switching the three focal modes with appropriate power weightings. The contour lines from inside to outside indicate TD equal to 300, 100, and 1 min, respectively.

We also tried to increase the number of sonication paths from four to eight, and the heating process was divided into two phases with a 20 s–20 s heating duration. In the first phase, the four transducers heated the PTV through the original four paths. After heating the PTV simultaneously for 20 s, the set of transducers was rotated 45° instantly with respect to the Z-axis and heating resumed for another 20 s.

After the heating was finished, the entire computational domain was cooled naturally with a constant water temperature of 37 °C. The thermal dose continued to accumulate during the cooling process, and the overall treatment was complete once the peak temperature of the entire domain

dropped below 43 °C. The peak temperature of the whole domain, which was supposed to occur within the PTV, the peak temperature of surrounding normal tissues (0.5 cm beyond the six planes of PTV cube in their normal directions), and the peak temperature of the rib tissue were taken for assessment. These three peak temperatures were recorded from the beginning of heating until the end of treatment.

II.C. Calculation of pressure fields (secondary source model)

To accurately calculate the acoustic pressure field for the model with multiple mediums including water, soft tissue, and rib, we employed the method described previously by Sun and Hynynen.²³ The approach treats each interface as a reirradiator, and by portioning the wave propagation path into a series of layers, the algorithm can be performed to calculate the pressure field in the medium of interest after consecutive applications of the Rayleigh integral,

TABLE I. Phase combination and computed acoustic power weighting for each focal mode.

Focal mode	Phase combination ($n=0, 1, 2, 3$)	Acoustic power weighting (W/cm ²)
Mode 0	(0°, 0°, 0°, 0°)	2.04
Mode 1	(0°, 90°, 180°, 270°)	5.93
Mode 2	(0°, 180°, 0°, 180°)	7.26

FIG. 2. Schematic diagram of the proposed heating strategy. (a) The breast was modeled as a hemisphere with a radius of 6 cm, the chest wall as a 1-cm-thick slab, and the ultrasound transducer was tilted 45°. (b) The breast was sonicated from four directions with the focal center of each sonication 6 mm away from the geometrical center of the planned target volume (PTV), and the sonication sequence was also indicated.

$$\begin{aligned}
 p = & \sum_{i_1}^{N_1} \dots \sum_{i_M}^{N_M} \prod_{l=1}^{M-1} \left[\frac{ik_{c_l} e^{-ik_{c_l} R_{i_l, i_{l+1}}}}{2\pi R_{i_l, i_{l+1}}} \right. \\
 & \times \left(1 - i \frac{1}{k_{c_l} R_{i_l, i_{l+1}}} \right) \cdot T_{v_{i_{l+1}}} \cos(\theta_{i_{l+1}, t}) \left. \right] \\
 & \times \left(\frac{ik_{c_M} \rho_M c_M e^{-ik_{c_M} R_{i_M}}}{2\pi R_{i_M}} \right) u_{i_l} \left(\prod_{l=1}^M ds_{i_l} \right), \quad (1)
 \end{aligned}$$

as presented in the formula, p is the overall pressure field, u_{i_l} is the normal velocity of the i_l th discrete point source belonging to the l th layer interface, and k_{c_l} denotes the complex wave number in layer l . $R_{i_l, i_{l+1}}$ represents the propagation distance between i_l th simple source of area ds_{i_l} on the l th layer's interface and the i_{l+1} th simple source of area $ds_{i_{l+1}}$ on the $l+1$ th layer's interface. The density and sound speed of the final layer are ρ_M and c_M , respectively, and $\theta_{i_{l+1}, t}$ denotes the transmission angle of the wave while it propagates

through the layer $l+1$. For the i_l th simple source on the l th layer interface, the particle velocity transmission coefficient T_{v_l} can be denoted by

$$T_{v_l} = \frac{2\rho_{l-1}c_{l-1}/\cos \theta_{i_l, i}}{(\rho_l c_l / \cos \theta_{i_l, t}) + (\rho_{l-1}c_{l-1} / \cos \theta_{i_l, i})} \cos \theta_{i_l, i}, \quad (2)$$

where $\theta_{i_l, i}$ and $\theta_{i_l, t}$ are incident and transmitted angles, respectively; they satisfy Snell's law. The absorbed acoustic power deposition, q (W/m^3), is given by²⁴

$$q = \frac{\alpha |p|^2}{\rho c}, \quad (3)$$

where ρ denotes the density of tissue (kg/m^3), c is the speed of sound in tissue (m/s), p is the pressure (Pa), and α is the pressure amplitude absorption coefficient (Np/m). In this study, the scattering effect is not considered, and therefore the absorption coefficient is assumed to be equal to the at-

TABLE II. Properties of tissues used in the simulation (Refs. 14 and 18).

	Acoustic parameters			Thermal parameters		
	Tissue density (kg/m^3)	Sound speed (m/s)	Attenuation coefficient (Np/m/MHz)	Thermal conductivity (W/m^2C)	Specific heat ($J/kg/^\circ C$)	Blood perfusion ($kg/m^3/s$)
Breast	1020	1553	8.68 ^{1.5*Frequency}	0.5	3770	1.7
Rib	1600	3400	200	0.4	1000	0.5

FIG. 3. (a) A prototype of the transducer in this study. (b) Focal pattern of mode 0 measured by hydrophone (normalized to peak value, unit: dB). (c) The set-up used for *in vitro* tissue model experiments in this study.

tenuation coefficient, which is taken into account by the imaginary part of the complex wave number $k_c = k - j\alpha$, where k is the real wave number ($2\pi/\lambda$, λ is the wavelength) in the medium. All kinds of tissues, including breast and rib, are assumed to be homogeneous with a uniform attenuation coefficient in pressure field calculations. The acoustic parameters of rib and breast tissues are listed in Table II,^{14,19} and the parameters of water were set at $c = 1519$ m/s, $\rho = 1000$ kg/m³, and $\alpha_r = 0$ Np/m.¹⁴

II.D. Temperature and thermal dose calculation

Pennes' transient bioheat transfer equation²⁵ was used to simulate the temperature distribution

for a given absorbed power deposition obtained from Eq. (3),

$$\rho c_t \frac{\partial T}{\partial t} = k_t \nabla^2 T - w_b c_b (T - T_{ar}) + q \quad (4)$$

where ρ is the density of tissue (kg/m³), c_t and c_b are the specific heat of tissue and blood (J/kg/°C), k_t is the thermal conductivity of tissue (W/m/°C), w_b is the perfusion rate (kg/m³/s), and T_{ar} is the arterial temperature (37 °C). The thermal properties used for breast and rib tissues were selected according to previous studies.^{26–28} The values used in simulation are listed in Table II, and the specific heat for blood was set at 3770 (J/kg/°C). A three-dimensional finite

FIG. 4. Simulation results for heating a PTV centered at (0,0,-2). (a–c) show the TD distributions on the $Z = -2$, $Y = 0$, and $X = 0$ cm planes respectively. The contour lines from inside to the outside indicate TD equal to 300, 100, and 1 min, respectively.

TABLE III. Detailed information of heating PTVs with various locations in simulations.

Center location of PTV	Peak temp. (°C)	Peak temperature in surrounding tissue(°C)	Peak temperature in rib (°C)	Treatment time (s)
(0,0,-2)	75.33	55.03	46.99	405.40
(0,0,-3)	77.67	56.23	43.04	412.15
(0,-3,-3)	78.71	56.51	43.34	407.45
(0,0,-4)	80.51	56.48	40.91	418.10

difference method was used to obtain the numerical solution of the temperature distribution. The spatial and temporal grid sizes used were 0.5 mm and 0.05 s, respectively, and the temperature of the water surrounding the breast was set to a constant value of 37 °C.

To calculate the thermal dose equivalent in minutes at 43 °C in order to estimate the extent of tissue necrosis caused by thermal damage, the Saparedo–Dewey expression,²⁹

$$TD = \int_{t_0}^{t_f} R^{(T-43)} dt = \sum_{t_0}^{t_f} R^{(T-43)} \Delta t, \quad (5)$$

was used, where $R=2$ for $T \geq 43$ °C and $R=4$ for 37 °C $< T < 43$ °C, t_0 and t_f are the initial and final time to calculate the thermal dose, respectively, Δt is the discrete time step, and TD is the thermal dose in units of time. The threshold dose of total necrosis ranges from 50 to 240 equivalent min at 43 °C for different tissues.^{30,31} In this study, a conservative value (TD=300 min) was taken as the threshold for tissue necrosis, and all the simulations were implemented using MATLAB®7.0 in a three-dimensional domain.

II.E. *In vitro* experiment

A corresponding prototype transducer, as shown in Fig. 3(a) (BroadSound Co., Hsinchu, Taiwan), was constructed to demonstrate the feasibility of the heating strategy. The pressure measurement using hydrophone for the focal plane of

mode 0 was conducted, and the resulting pattern is shown in Fig. 3(b). The main components of the experimental set-up are presented in Fig. 3(c), in which the transducer was driven by a four-channel power amplifier (500-012 RF Generator, Advanced Surgical System, Inc., USA). The power amplifier was controlled by the software installed on a personal computer. The applied power and phase for each element of the transducer could be controlled. A fresh porcine muscle, used as an *in vitro* tissue model, was placed in a 37 °C degassed water bath and allowed to reach temperature equilibrium before heating. The tissue in the experiment was placed on top of a cylindrical water tank filled with 37 °C degassed water, and the transducer was tilted at an angle of 45° and placed on a stand driven by a controllable motor at the tank bottom. By programming the motor controller, the transducer could be rotated to four predetermined positions in the desired sequence.

III. RESULTS

III.A. Simulation results

III.A.1. Single-transducer strategy

Figure 4 shows the resulting thermal dose profiles on the $Z=-2$, $X=0$, and $Y=0$ cm planes for the PTV centered at (0,0,-2) after treatment was completed using the single-transducer strategy. The contour curves from the inside to the outside indicate that TD is equal to 300, 100, and 1 min. The

FIG. 5. Simulation results for heating a PTV located at different locations. (a–c) show the TD distributions on the $x=0$ cm plane for the PTV centered at (0,0,-3), (0,-3,-3), and (0,0,-4), respectively.

FIG. 6. Simulation results for heating a PTV centered at (0,0,-2) with different heating strategies. (a) a single ultrasound transducer sonicated from four directions sequentially (total heating time=40 s); (b) four ultrasound transducers sonicated from four directions simultaneously; (c) four ultrasound transducers sonicated from four directions simultaneously, followed by 45° rotation of the transducers for a second set of four sonications.

heating duration was 8 s for each sonication, and the total sonication time was 32 s for the entire PTV. The total treatment time, which is defined as the time required for the peak temperature to drop below 43 °C, was 405.4 s. The details of the simulation results are listed in Table III. It was observed that the 300 min TD contour covered the PTV adequately, and even though there was no cooling interval between heating phases, the peak temperature of the surrounding tissues was still about 55 °C, and the peak temperature of the ribs was about 47 °C for the entire treatment.

To evaluate the flexibility of the single-transducer strategy for treating PTV in different locations, the same heating procedure and control parameters were also simulated for the PTV located at (0,0,-3), (0,-3,-3), and (0,0,-4). The resulting thermal dose contours on the $X=0$ plane for these three PTV locations are shown in Figs. 5(a)–5(c) and the details of simulations are listed in Table III. It was observed that TD=300 min contours are consistent with those in the previous case. Since the focal zone is aimed at the PTV, the peak temperature of the ribs increases as the PTV approaches the ribs. However, they were all below 47 °C, as shown in Table III. Conversely, the peak temperature of the breast tissue decreased due to a longer penetration depth.

The TD profiles heated by the single-transducer strategy without heating time modification are shown in Fig. 6(a) for a comparison of three strategies. It was found that when the heating duration for each unit was maintained for 10 s instead of 8 s, in contrast to Fig. 4(c), the excessive TD=300 min appeared in surrounding tissue.

III.A.2. Four-transducer strategy

Figure 6(b) shows the TD profiles on the $X=0$ cm plane with four transducers driven simultaneously to heat a PTV centered at (0,0,-2). It can be seen that the PTV was completely covered with the contour of TD=300 min, and there is no TD higher than 100 min in surrounding normal tissue and rib. The TD profile on the $X=0$ cm plane produced by the strategy integrating four transducers with mechanical rotation is illustrated in Fig. 6(c), and the TD accumulation tends to be more concentrated on the PTV region treated by this eight-path heating strategy.

III.B. In vitro experiments

An equal duration for all focal modes with the order: Mode 0 → Mode 1 → Mode 2 was used in *in vitro* experiments. Mode 2 was driven with an electrical power of 200 W, which is the power limitation of the prototype transducer. Based on the acoustic power weighting ratio in simulations, as shown in Table I, the proportional powers were applied to the other two modes, as shown in Table IV. The heating duration for each mode was 26 s, producing a heating unit of ablation volume approximately equal to that of the simulation case ($1 \times 1 \times 2$ cm³).

Following the sonications of four different paths accompanied by three rotation intervals (about 10 s for each), the tissue phantom was sliced through the center of the lesion, and its characteristic length was measured and photographed.

TABLE IV. Parameters of each mode used to form a heating unit in experiments.

Focal mode	Electrical power (W)	Heating duration (s)
Mode 0	56	26
Mode 1	164	26
Mode 2	200	26

FIG. 7. Thermal lesion on X-Y and X-Z planes, produced by single transducer with multidirectional heating.

FIG. 8. Simulation results for heating a PTV centered at (0,0,-2) using an experimental heating method. (a-c) TD distributions on $Z=-2$, $Y=0$, and $X=0$ cm plane, respectively.

Two representative cross-sectional photos, on the X - Y and X - Z planes, are illustrated in Figs. 7(a) and 7(b), respectively.

IV. DISCUSSION

The referenced method²² for TD optimization is aimed at forming an ideal distribution of TD within a single heating unit. Therefore, if the optimized heating unit is applied consecutively to treat the entire PTV, the excessive TD in the surrounding tissue would occur as shown in Fig. 6(a). To obtain a better TD distribution for the entire treatment region, a secondary optimization was conducted by shortening the heating duration for each heating unit (from 10 to 8 s).

Figures 4 and 5 show the simulation results of a single transducer with multidirectional heating for a $2 \times 2 \times 2$ cm³ PTV located at different locations. The results indicate the single-transducer strategy is capable of producing an appropriately conformal TD distribution for PTVs located at different locations in the breast, and the TD accumulations in surrounding normal tissues and ribs were controlled properly. The peak temperature in the surrounding tissue for all the cases mentioned before were all slightly higher than 56 °C (Table III), resulting in a small region of TD higher than 100 min in the surrounding tissue.

The parameters used in the simulations were not exactly implemented in the *in vitro* experiment; however, the coagulation pattern produced in experiment (Fig. 7) still provided a direct verification for the simulated TD patterns predicted by the multidirectional heating scheme. For detailed examination, a simulation case using the heating method in the experiment was also conducted in this study. To simulate the

experimental condition, a simulation study was conducted to heat a PTV with heating duration of 78 s and 0.128 times of power level in Table I for each heating unit. The resulting TD profiles are shown in Fig. 8, and by comparing Figs. 8(a) and 8(b) with the thermal lesions shown in Figs. 7(a) and 7(b), an observable similarity in size and pattern between the simulation and experiment is demonstrated. Furthermore, in contrast to the ideal patterns produced by 10 Hz switching [Figs. 4 and 6(a)], the region covered by TD=300 min in simulation using experimental conditions is smaller and asymmetrical. This phenomenon implied that when the applied total energy is the same, a sonication with lower power and a longer heating duration would result in a smaller lesion and the patterns of TD profiles would be more affected by heat conduction due to a slower temperature rise. Moreover, considering the breast structure, if the focal zone encounters the heterogeneity with large variations in blood perfusion rate and/or acoustic absorption, the heterogeneity may cause uneven distribution of the TD to induce the phenomenon of hot and/or cold spots.

On the basis of identical energy delivered and heating duration, when we compared Fig. 6(b) with Fig. 6(a), it was observed that the TD profiles of Fig. 6(b) were more symmetrical since tissues on the four paths were heated from the same initial temperature level, and unlike the excessive TD = 300 min found outside the PTV region in Fig. 6(a), there is no TD = 100 min in the surrounding normal tissue in Fig. 6(b). As for the strategy integrating four transducers with rotation, the heating duration in normal tissue was half that of the previous case without rotation, due to the increase of

TABLE V. Comparison of heating PTV centered at (0,0,-2) with different strategies in simulations.

Heating strategy	Peak temperature (°C)	Peak temperature in surrounding tissue (°C)	Peak temperature in rib (°C)	Treatment time (s)
Single-transducer	82.91	56.74	48.49	477.10
Four-transducer	81.37	51.80	46.10	476.85
Four-transducer and rotation	78.33	50.06	45.96	463.35

FIG. 9. The responses of peak temperatures in the PTV, surrounding tissue of PTV, and rib tissue for different sonication strategies. (a) Using a single transducer with sonications from four different directions sequentially, (b) using four transducers sonicated from four directions simultaneously, and (c) using four transducers sonicated from four directions simultaneously, followed by four sonications from transducers rotated 45° .

path number for sonications. Therefore, the TD in surrounding normal tissue is even lower in Fig. 6(c) than that in Fig. 6(b). Detailed comparison of the simulation using the three strategies is listed in Table V. To further contrast the differences among the three strategies, the responses of peak temperatures in the PTV, the surrounding normal tissue, and the ribs in simulations are presented in Fig. 9. The three sets of peak temperature records are illustrated in the order of corresponding cases shown in Figs. 6(a)–6(c). Figure 9(a) shows the temperature response for a single transducer sonicating sequentially from four different directions, and reveals that three peak temperatures dropped during the initial stage of the next phase of sonication, although no cooling interval was inserted between sonication. This indicates that with path shifting, the location of the peak temperature changed with the shift of sonication path, and it appeared sequentially in the focal zone of each acoustic beam, as shown in Fig. 2(b). Thus, as shown in Table V, the localized heating in the surrounding tissue was inhibited below 57°C , and the rib temperature was kept under 49°C throughout the heating phase, yet the targeted tissue within the PTV was heated to 82°C due to the convergence of the acoustic beams. Nevertheless, for a temperature response in the surrounding tissue, transient high temperatures above 55°C is still regarded as

harmful and responsible for the excessive TD=300 min, as shown in Fig. 6(a). Since the same amount of total energy for four heating units was divided by four in the four-transducer strategies, the spatial power intensity in these strategies is one-quarter of that in the single-transducer strategy. Therefore, lower spatial power deposition in the second and third cases decreased the rise of peak temperatures. As shown in Figs. 9(b) and 9(c), the peak temperatures of the PTV in these two cases are both lower than 82°C , and the peak temperatures in the surrounding tissue and rib are below 52°C and 47°C , respectively. As we compare these two strategies both utilizing four transducers, it was found that peak temperatures of the surrounding tissue and ribs in the eight-path strategy [Fig. 9(c)] are even lower, below 51°C and 46°C , respectively. Thus, the heating of surrounding normal tissue was further alleviated by increasing the sonication paths from four to eight. Additionally, the locations of four peak temperatures produced simultaneously by four focal zones were located, as shown in Fig. 2(b), while they shifted 45° in the four-transducer with mechanical rotation strategy.

Even though the entire temperature alleviation of the ribs was not experimentally verified in this study, from Fig. 7(b), it is shown that the tissue above the PTV maintains the origi-

nal color. Furthermore, in all simulation cases, as shown in Fig. 6, there is only a small portion of ribs with TD > 1 min for the three heating strategies. Meanwhile, the peak temperatures of the ribs shown in Fig. 9(a) only transiently approached 48 °C, and in the other two strategies, they were both below 47 °C. Therefore, considering the proper TD distribution and low temperature rising in the ribs, a 45° angle for the transducer is large enough to sufficiently alleviate rib heating.

The scattering effect was not considered in the simulation study and the attenuated energy was assumed to be 100% transferred to absorption energy. However, in a real situation, both coefficients increase with temperature rise and TD accumulation during heating, and only about 60% to 85% of attenuated energy would be transferred to effective absorption.³² Therefore, because the temperature and thermal dose are much higher in the PTV than in the surrounding normal tissue, it can be inferred that the temperature rise in the PTV would be higher than the predicted value in simulation, while it would be lower in the surrounding normal tissue.

In all simulated cases based on the multidirectional heating scheme, whether by single or four-transducer strategies, the treatment times for a PTV of volume $2 \times 2 \times 2 \text{ cm}^3$ were all less than 8 min. Even in the *in vitro* experiment completed by single transducer, the heating of a thermal lesion of volume $2 \times 2 \times 2 \text{ cm}^3$ can still be finished within 8 min. The ablation rate (ablated volume/time) is much faster than reported previously (an average of 1.3 h for tumors with a mean size of 3.1 cm in diam.), and the strategies with four-transducer strategies are potentially much faster.

V. CONCLUSION

The aims of this study were to effectively shorten the entire treatment time and to alleviate rib overheating during high-intensity focused ultrasound for breast tumor thermal surgery. To achieve these goals, this study used a single split-focused ultrasound transducer or multiple split-focused transducers with mechanical rotation to do multidirectional heating. The transducer was driven by a focal switching technique and tilted to 45° to induce total reflection at the muscle/bone interface. Using a single transducer, a $2 \times 2 \times 2 \text{ cm}^3$ PTV can be appropriately treated without cooling intervals, while effectively controlling the heating of the ribs and surrounding normal tissue. In addition, from the simulations with four transducers, it can be inferred that by using multiple transducers to heat PTV simultaneously, or increasing the number of sonication directions, the performance of the multidirectional scheme could be further improved. The results of this study indicate that the system is very promising to shorten the entire treatment time and alleviate rib overheating while conducting focused ultrasound thermal surgery for breast tumors.

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