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Investigation of a Cylindrical Ultrasound Phased-Array with Multiple-Focus Scanning for Breast Tumor Thermal Therapy

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Abstract— **The purpose of this study is to investigate the feasibility of a cylindrical ultrasound phased-array with multiple-focus scanning strategy to produce a uniform heating for breast thermal therapy. In this study, a breast is surround by a 1-MHz cylindrical ultrasound phased-array consists of 200 elements with a radius of 10 cm and a height of 2 cm. To prevent overheating in the normal tissue, a scanning region of 1㎝×1㎝ was selected as a single heating unit. Planning target volume (PTV) larger than this size would be divided into several sub-heating units, and then be treated sequentially with cooling phase to prevent overheating in the surrounding normal tissue. Parameters such as the target temperature, blood perfusion rate and the size of PTV are evaluated. Simulation results show that the target temperature affects the thermal lesion size and the blood perfusion rate increases the heating time significantly. This method provides efficient heating for breast tumor thermal therapy while preventing overheating the ribs.**

Keywords: cylindrical ultrasound phased-array, multiple-focus scanning, breast tumor, thermal therapy

I. INTRODUCTION

A truly non-invasive treatment method, ultrasound thermal therapy, is useful for tumor ablation deep in the body. It can precisely ablate the tumor when accumulating thermal dose higher than a critical level. The breast is particularly well suited for ultrasound heating due to the low blood perfusion and the convex shape which provides a sufficient acoustical window for delivering ultrasonic energy. The preliminary studies of ultrasound thermal therapy for treatment of breast tumors using single element transducer [1] or the hemispherical phased-array transducer [2] have been accomplished. The main limitation of a single element transducer is the small focal size. Therefore, the treatment time is significantly increased when the tumor is large. A recent clinical report shows that it takes up to 240 minutes for treating tumors with a mean size of 22 mm in diameter [3]. Hemispherical phased-array has shown limitations when the tumor is close to ribs due to the high absorption coefficient of ultrasound in ribs. The acoustic absorption in ribs at frequencies of 0.5 to 5 MHz generally used for the ultrasound thermal therapy is one or two order(s) of magnitude greater than that in the soft tissue [4]. It may cause the limitation of the treatment such as the patient pain by generating localized high temperature.

To prevent overheating in ribs, the ultrasound power deposition in ribs must be minimized. To achieve this goal, a cylindrical ultrasound phased-array surrounding to the breast was proposed. The emitted ultrasound beam from this cylindrical phased-array approximately paralleled to the ribs; hence results in a total reflection of ultrasound energy at the rib-breast interface. Therefore, not only the power deposition but also the temperature rise of ribs can be effectively minimized. The preliminary simulations were performed to investigate the feasibility of this cylindrical ultrasound phased-array for producing an efficient heating for breast tumor thermal therapy while preventing overheating the ribs.

II. METHODOLOGY

A. System configuration

A schematic diagram of the proposed system is illustrated in Fig.1. A 1-MHz cylindrical ultrasound phased-array consisted of 200 elements with a radius of 10 cm and a height of 2 cm. The breast was modeled as a 6 cm-radius hemisphere and was surrounded by the phased-array. The geometrical center of the cylindrical ultrasound phased-array was mechanically moved to the center of the desired treatment region. Water was filled between the phased-array and breast as the coupling medium. The main advantage of this arrangement is that the ultrasound beam path from the phased-array is approximately parallels to the ribs. Besides, the acoustic window was used to alleviate the near-field overheating efficiently. Therefore, not only the power deposition but also the temperature rise of ribs and the near-field overheating could be minimized.

B. Acoustic model

The acoustic model was a water-tissue model. The density and the speed of sound in each two medium are very close, the reflection and refraction at the water-tissue interface could be ignored [5]. Continuous wave sonication was used in the simulation. The ultrasonic pressure field was calculated by the Rayleigh-Sommerfeld to integrate the contribution of each point source on the surface of the phased-array. Parameters used in simulation are listed in Table I [6], [7]. The driving signals for the phased-array elements that produce a specific focused pattern are calculated by using a pseudo inverse method with a limited maximum output power intensity of 5W/cm^2 for the phased-array elements.

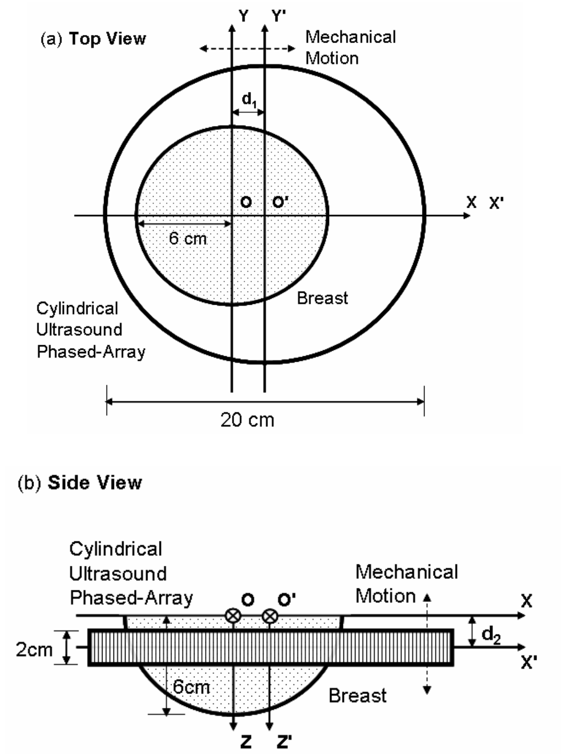

Fig. 1. Schematic diagram of the cylindrical ultrasound phased-array and the breast: (a) top view (b) side view.

C. Thermal model

The temperature response in tissue induced by thermal conduction, blood perfusion and absorbed ultrasound power deposition were modeled by using the Pennes bioheat transfer equation [8]. The thermal dose (TD), represents the equivalent minutes at 43℃ that used to estimate the tissue necrosis, was calculated numerically by using Sapareto and Dewey's thermal dose function[9]. Tissue was considered to be necrotic when TD exceeded 240 min.[10].

TABLE I
ACOUSTIC AND THERMAL PROPERTIES OF BREAST TISSUE USED IN SIMULATION

Symbol	Quantity	Value
ρ	tissue density	1000（kg/m^3）
c	speed of sound	1500（m/s）
α	ultrasonic absorption coefficient	5（Np/m/MHz）
k	thermal conductivity	0.5（W/m/℃）
c_t / c_b	specific heat (tissue/blood)	3770（J/kg/℃）
W_b	blood perfusion	0.5（$\text{kg/m}^3\text{/s}$）

D. The heating strategy

Comparing to a single-focus for each sonication, the multiple-focus method enlarges the focal region while reducing the requirements of large number of sonications when the desired heating region is large. Thus, multiple-focus patterns were used for heating in this study. Fig.2. shows the arrangement of multiple-focus patterns. Each pattern consists of four foci with 0.2cm apart. To produce a conformal heating, the predetermined multiple-focus patterns were uniformly distributed within the scanning region in X-Y plane (the treatment plane). The center of each multiple focus pattern, was used as a temperature control point (TCP). During the heating phase, multiple-focus patterns were sequentially and repeatedly scanned with a 1s sonication time at each predetermined position. No cooling phase was used between sonication. As the temperature of the TCP reached the target temperature (T_{tgt}), the corresponding multiple-focus pattern would not be sonicated for the following scanning. The scanning heating phase stopped when all TCPs had ever reached the T_{tgt}.

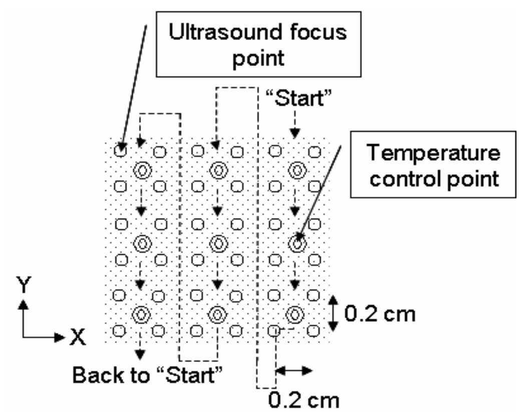

Fig. 2. Scheme of multiple-focus scanning and the locations of temperature control points.

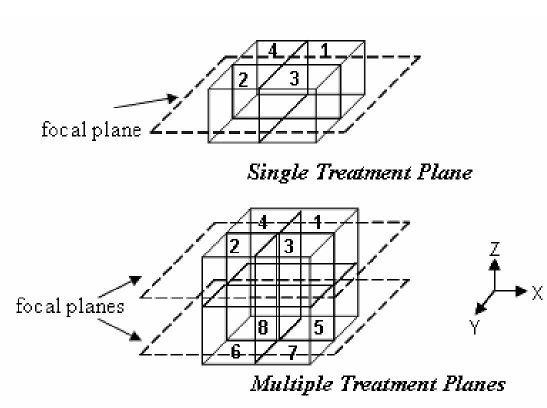

Fig. 3. Arrangement of single (upper) and multiple (lower) treatment planes.

During the scanning heating process, the acoustic wave is absorbed along the propagation path. When the scanning range is large, it may result in an overheating of the surrounding normal tissue. Hence, a scanning heating region of 1㎝×1㎝ on the treatment plane was selected to be a heating unit. PTV larger than this size would be divided into several sub-heating units and then heated sequentially. An appropriate cooling phase was needed between sub-heating unit treatments. Fig.3. shows the arrangement of the treatment planning for large PTV. The PTV is divided into several sub-heating units and the number 1 to 8 indicates the heating sequence.

The thickness of the resulting heating pattern is dependent on the size of the focus in Z-direction. The size of -6dB intensity of focus in Z-direction (0.8㎝ to 1㎝) is a good approximation of the thickness of the resulting heating pattern. If the PTV is large in Z-direction, the PTV was divided into several treatment planes and then heated separately to cover the whole PTV.

III. Results and Discussion

To evaluate the influence of T_{tgt} on the treatment, three T_{tgt} (54, 56 and 58℃) were examined. Fig. 4 shows the heating results for a 1㎝×1㎝ heating unit at (0, 0,-1) for T_{tgt} of 54, 56 and 58°C (red, green and blue lines, respectively) with blood perfusion rate 0.5kg/m^3/s. Fig. 4(a)-4(c) indicate the TD240 min. contour on Z=-1, X=0 and Y=0 plane, respectively. Results are summarized in TableⅡ. Obviously, the lesion size increases with the target temperature, T_{tgt}. It is sufficient to produce a 1㎝×1㎝ lesion on the treatment plane when T_{tgt} is higher than 54℃. The thickness (in Z-direction) of the lesion is thin when T_{tgt} is low; while, lesion size can be extended on the treatment plane with a higher T_{tgt}. For an accurate conformal heating, low T_{tgt} should be used. And more treatment planes are needed when the PTV is large in Z-direction. T_{tgt} = 56℃ was used for the following simulations.

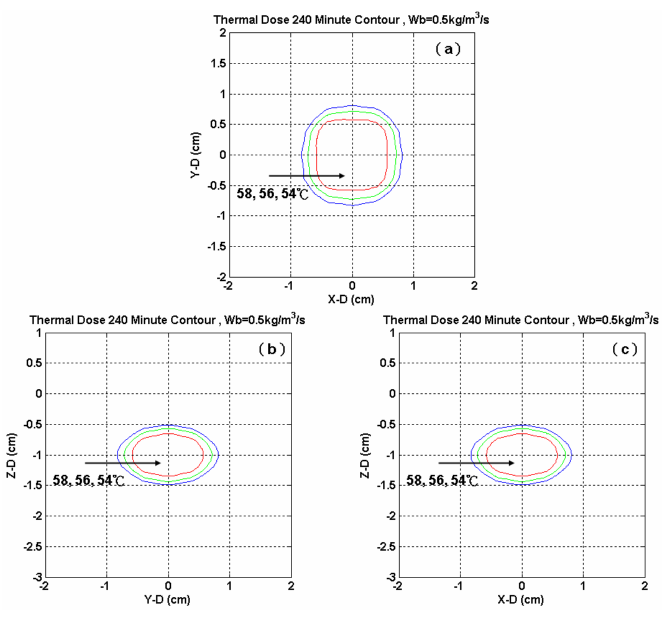

Fig. 4. Heating results of a single heating unit centered at (0,0,-1) with T_{tgt} 54, 56, or 58℃ and TD240 min. contour on (a) Z= -1 (b) X=0 and (c) Y=0 plane. The blue, green and red line indicate the T_{tgt}=58, 56 and 54℃, respectively.

Traditional heating strategy is to completely ablate tumor tissue around the focal spot by a single sonication in few seconds, then wait for cooling to accumulate the thermal dose to produce a small lesion. A large lesion can be integrated by a set of small lesions. Thus, the effect of blood perfusion is not significant. In this study, the proposed heating strategy is to scan the focus through out the predetermined scanning region. Generally, it takes more than one minute to complete a treatment for a size of 1㎝×1㎝. The effect of blood perfusion becomes an important factor at high blood perfusion level. Fig.5. shows the effect of blood perfusion on thermal dose distributions. In this simulation, the heating unit was 1cm×1cm centered at (0, 0,-1) while T_{tgt} was 56 ℃. Three blood perfusion rates 0.5, 5 and 10 kg/m^3/s were examined. Results are summarized in TableⅡ. Fig. 5(a) and 5(b) display the contours of thermal dose higher than 240 minutes, which is defined as the thermal lesion, on treatment plane. It can be seen that there is no obvious difference among the lesion sizes for blood perfusion rates. Figures 5(a') and 5(b') further demonstrate the thermal dose distribution along the X and Z direction, respectively. The horizontal dashed line indicates TD=240 min. and the vertical dashed lines indicate the boundary of heating unit.

Table Ⅱ

Simulation Results of the Single Heating Unit

W_b ($kg\ m^{-3}s^{-1}$)	0.5			5	10
T_{tgt} (℃)	54	56	58	56	
Sonication time (sec.)	65	77	87	93	151
Max. temperature (℃)	58.2	60.5	62.7	59.6	58.7

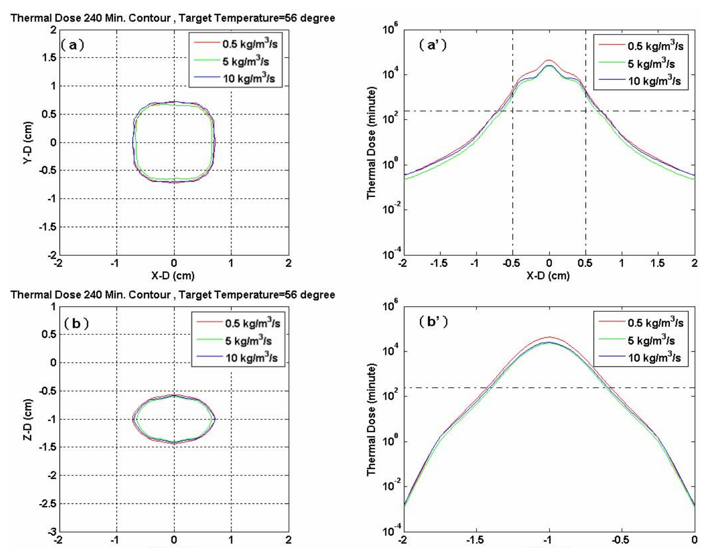

Fig. 5. Heating results of a single heating unit at (0,0,-1) for W_b=0.5, 5 and 10 kg /m^3/s. TD 240 min. outlines on (a) Z= -1 (b) Y= 0 plane. Thermal dose distribution on (a') X-axis (b') Z-axis.

It was observed that the maximum temperature is always at the center of the heating unit. For a lower blood perfusion rate, the maximum temperature is higher (Table II); and, results in a higher thermal dose (Fig. 5(a') and 5(b')). However, on the boundary of the heating unit, the 240min. thermal dose contours converge together for different perfusion rates.

To investigate the feasibility of the proposed system for heating a large tumor, Fig. 6 shows the simulation results for a PTV with a 2㎝×2㎝ cross-sectional area centered at (3, 0,

-1) with T_{tgt}=56℃ and W_b=0.5 kg/m^3/s. The PTV was divided into four sub-heating units, each sub-heating unit is 1cm×1cm, as shown in Fig. 6(b). The number indicates the heating sequence. There is a cooling phase following each sub-heating unit treatment until the maximum temperature lower than 43℃. The heating and cooling times required for each sub-heating unit treatment are summarized in Table Ⅲ. It takes about 1 minute to heat a sub-heating unit, except the unit 1. The total treatment time (from the beginning of heating sub-unit 1 to the end of heating sub-unit 4) was 835 seconds.

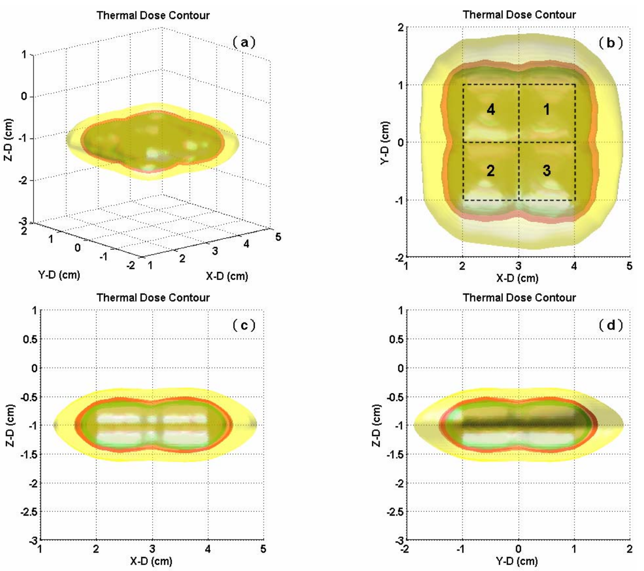

Fig. 6. Isosurface of TD for large PTV treatment (a) 3D view (b) top view (c) side view from right side (d) side view from left side

TABLE Ⅲ
SIMULATION RESULTS FOR A 2CM*2CM PTV

	Sub-unit 1	Sub-unit 2	Sub-unit 3	Sub-unit 4
Sonication time (sec.)	75	62	57	61
Cooling time (sec.)	184	198	198	197

IV. CONCLUSION

This study investigated the feasibility of a cylindrical ultrasound phased-array for producing a uniform heating in breast tumor thermal therapy. Multiple-focus pattern was generated by phased-array to enlarge the focal zone in each sonication. The target temperature, T_{tgt}, affects the resulting lesion size. The heating time increases with the blood perfusion rate significantly; however, its influence on the thermal lesion size is not obvious. The main advantage of this study is that the proposed system may treat a large tumor efficiently without near-field overheating due to large acoustic window. Furthermore, the ultrasound beam path emitted from the cylindrical ultrasound phased-array is approximately parallel to the ribs; hence it is capable of preventing the undesired temperature rise in the ribs during the treatment

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