Published in final edited form as:

Ultrasound Med Biol. 2010 October ; 36(10): 1704–1712. doi:10.1016/j.ultrasmedbio.2010.07.018.

Theoretical and experimental study on temperature elevation behind ribs caused by weakly focused ultrasound

Xiaozhou Liu^{*}, Chang Yin^{*}, Xiufen Gong^{*}, and Wenwu Cao[‡]

^{*}Key Laboratory of Modern Acoustics, Institute of Acoustics, Nanjing University, Nanjing 210093, People's Republic of China

[‡]Material Research Institute, The Pennsylvania State University, University Park, PA 16802

Abstract

Temperature distribution in tissues behind ribs produced by **weakly** focused ultrasound had been calculated using Pennes bioheat equation, and the validity of the theoretical model was experimentally confirmed in vitro using porcine live. We found that the position of the maximum temperature in tissue is strongly influenced by the distance between the transducer and ribs, while the gap between ribs is the determining factor for the maximum achievable temperature. Within the focal length, when the distance between the transducer and ribs increases, the maximum temperature **increases** and its position shifts away from the transducer. The rib width has little effect on the position of the maximum temperature but affects the achievable peak temperature. Our results provide useful information for treating liver cancers using ultrasound **induced hyperthermia**.

Keywords

focused ultrasound; ribs; hyperthermia; Pennes bioheat equation

INTRODUCTION

Ultrasound can penetrate into the interior of human body for clinical treatments. Many ultrasound-based therapeutic methods have been developed in the past 30 years. Some of them, such as ultrasound induced hyperthermia (Diederich et al. 1999; Matsumine et al. 2007) and high-intensity focused ultrasound (Chaussy et al. 2005; ter Haar et al. 2007; Wu et al 2004; Li et al 2004; Gunnar et al. 2006), are already used clinically for cancer treatment. Although ultrasound has great potential for therapeutic treatments, there are some limitations preventing it **for certain applications** (Chavrier et al. 2000; Illing et al. 2005; Wu et al. 2004), for example, the presence of bones in the beam pathway is a major problem in ultrasonic thermal therapy. The acoustic absorption of sound waves in bones at frequencies commonly used for hyperthermia is more than an order of magnitude higher than that in soft tissues (Goss, 1978). In addition, acoustic waves are strongly reflected by **bones** so that part of the acoustic energy will be blocked. Higher acoustic absorption and strong reflection caused by bones affect the

© 2010 World Federation for Ultrasound in Medicine and Biology. Published by Elsevier Inc. All rights reserved.

Corresponding author: Prof. Wenwu Cao, 164 Materials Research Lab., The Pennsylvania State University, University Park, PA 16802, Phone: (814) 865-4101, Fax: (814) 865-2326, dzk@psu.edu.

Publisher's Disclaimer: This is a PDF file of an unedited manuscript that has been accepted for publication. As a service to our customers we are providing this early version of the manuscript. The manuscript will undergo copyediting, typesetting, and review of the resulting proof before it is published in its final citable form. Please note that during the production process errors may be discovered which could affect the content, and all legal disclaimers that apply to the journal pertain.

temperature distribution in tissues and produce uncertainties in ultrasound treatments. Therefore, it is important to develop a practical theoretical model addressing the blocking effect of bones to provide guidance for therapeutic applications of ultrasound induced hyperthermia.

The temperature distribution produced by ultrasound may be obtained by solving the Pennes bioheat equation (Pennes 1948) for most situations. For example, using a finite-difference method Qian *et al.* studied the steady state temperature field in the human body generated by concave focused ultrasound transducer for hyperthermia (Qian et al. 2001). In their article, the influence of parameters and tissue properties on the effective treated area was analyzed by introducing the concepts of heat focal distance and isotherms. Wu *et al.* calculated the steady state temperature elevation along the beam axis generated by a focused Gaussian beam (Wu et al. 1990). They found that the temperature rise is an increasing function of the ultrasound intensity but not significantly affected by the focal point adjustment at the transducer-medium interface. They also discussed temperature **distribution** at the tissue-bone interface. Moros *et al.* used three different techniques to measure the temperature distribution in tissues behind bones (Moros et al. 2004). Fujii et al. **also** studied the temperature rise at the muscle-bone interface during ultrasound hyperthermia (Fujii et al. 1990).

In order to decrease the influence of bones in ultrasonic therapy, a time-reversal method by focusing pulsed ultrasonic waves through absorbing and aberrating layers was proposed to administer targeted treatment of brain tumors (Tanter et al. 1998; Aubry et al. 2008). Two-step hybrid virtual array-ray technique for beam focusing through the rib cage was proposed by Botros *et al.* (Botros et al. 1998). In addition, the feasibility of using a spherical ultrasound phased array for transrib liver-tumor thermal ablation was also investigated (Liu et al. 2007).

In order to improve the effectiveness of the ultrasound treatment, it is important to know the temperature field behind ribs. In this study, we investigated **theoretically and experimentally** the influence of rib cage parameters **on** the distribution of temperature field in tissues behind the ribs. A 3-D model was difference formulated using the alternating direction implicit (ADI) backward finite method (Lapidus et al. 1982) in combination with a prediction-refinement method. We also performed experiments to verify the validity of our **theoretical approach and theoretical results**. After the validation, the theoretical model was used to numerically **predict** the influence of several parameters to the temperature distribution **behind ribs**, including the distance between the transducer and ribs, rib width and gap size between ribs. The purpose of this study is to provide a theoretical approach to guide the clinical usage of focused ultrasound in treating liver cancer via hyperthermia.

Aubry et al used a high power 200-element phased array to investigate the influence of ribs on the HIFU treatment. The defocusing effect of the ribs was experimentally measured *in vitro* and a time reversal technique was introduced in order to decrease this effect. It was demonstrated that the temperature of the rib cage can be dramatically reduced by the time-reversal technique. Our approach differs from those earlier studies. We simulated the 3-D temperature distribution near the focal point of the transducer and also inside the ribs when the ribs are present. The theoretical model, which was validated by our experiments and the experimental data of Fujii et al. (1999) was used to numerically predict the influence of different parameters to the temperature field, including the distance between the transducer and ribs, rib width and the gap size between ribs.

METHOD

Theoretical model and numerical algorithm

The well-known Khokhlov–Zabolotskaya–Kuznetsov (KZK) equation (Kuznetsov 1970) was used to model the nonlinear wave propagation in water or biological tissues. Its dimensionless form can be written as follows:

$$\frac{\partial}{\partial T} \left(\frac{\partial P}{\partial Z} - NP \frac{\partial P}{\partial T} - A \frac{\partial^2 P}{\partial T^2} \right) = \frac{1}{4G} \Delta_{\perp} P, \quad (1)$$

where $P = \frac{p}{p_0}$ is the normalized pressure, in which p is the sound pressure and p_0 is the sound

pressure at the surface of the transmitter; $\nabla_{\perp} = \frac{\partial^2}{\partial X^2} + \frac{\partial^2}{\partial Y^2}$ is the dimensionless transverse

Laplace operator with respect to X and Y , where $X = \frac{x}{a}$, $Y = \frac{y}{a}$ and $Z = \frac{z}{D}$ are normalized coordinates; here a is the radius of the transmitter, D is the geometrical focal length; $T = \omega \tau$

is the retarded time, where ω is the angular frequency and $\tau = t - z / c_0$; $N = \frac{D}{Z_n}$ with

$Z_n = \frac{\rho_0 c_0^3}{\omega_0 \beta p_0}$ being the shock wave formation distance of a plane wave with angular frequency ω_0 , c_0 is the sound speed, ρ_0 is the density of the medium, β is the acoustic nonlinearity

coefficient; $A = \alpha D$ with $\alpha = \frac{\omega_0^2 b}{2 \rho_0 c_0^3}$ being the absorption coefficient, b is the dissipative

parameter; $G = \frac{Z_d}{D}$ with $Z_d = \frac{\omega a^2}{2 c_0}$ being the Rayleigh distance.

The sound pressure in the form of a complex Fourier series expansion is solved by means of the aforementioned ADI backward finite difference method combined with a prediction-refinement method. (Li et al. 2007).

Since the pressure of each harmonic is calculated, the total intensity can be calculated by:

$$I = I_0 \cdot 4 \sum_{n=1}^{N_{\max}} |C_n|^2, \quad I_0 = \frac{p_0^2}{2 \rho_0 c_0}, \quad (5)$$

where $N_{\max}$ is the maximum harmonics used in the calculation. In this work, we set $N_{\max} = 7$, a value which would not cause any appreciable energy reflection from the truncated harmonics to lower harmonic components in the **numerical** calculation. This choice of 7 also reflects the experimental situation. The bandwidth of our hydrophone is from 1 MHz to 10 MHz. We can not **detect** the eighth harmonics and beyond, although the 8th harmonics (1.13*8=9.04 MHz) is within the bandwidth of the hydrophone. **In other words**, the maximum **detectable** harmonics in the experimental is $N_{\max} = 7$.

The absorption of acoustic energy in the media can be expressed as:

$$Q_v = Q_{v0} \cdot 4 \sum_{n=1}^{N_{\max}} n^{\mu} |C_n|^2, \quad Q_{v0} = 2 \alpha I_0 \quad (6)$$

where $\mu = 1.1-1.3$ for biological tissues and $\mu = 1$ for ribs. The temperature distribution in the tissue can be **calculated** by the Pennes bioheat equation (Pennes 1948):

$$\rho_s C_s \frac{\partial T}{\partial t} = K_s \nabla^2 T - W_b C_b (T - T_\infty) + Q_m + Q_v, \quad (7)$$

where ρ_s is the tissue density, t is time, C_s is the specific heat of tissue, K_s is the tissue thermal conductivity, W_b is the blood perfusion rate, C_b is the specific heat of blood, T is the temperature of tissue, T_∞ is the temperature of blood, Q_m is the rate of heat generation per tissue unit volume, Q_v is the rate of heat generation by external heat source.

The parameters of liver used in our calculations are: $\rho_s = 1224(\text{kg} / \text{m}^3)$, $c_s = 1614(\text{m} / \text{s})$, $\alpha_s = 57.1 \text{Np} / \text{m}$, $\beta_s = 4.45$, $K_s = 0.5 \text{J} / (\text{W} \cdot ^\circ \text{C})$ and $C_s = 3550 \text{J} / (\text{kg} \cdot ^\circ \text{C})$. The sound intensity on the the surface of the transducer was $1.0 \text{W} / \text{cm}^2$. The acoustic power was measured by radiation force method (Hill 1970) and the sound intensity was calculated as the acoustic output power divided by the surface area of the transducer. The initial pressure amplitude is 0.2MPa .

For simplicity, the ribs are assumed to have a rectangular cross-section extending along the Y-axis. Hence, the sound beam is parallel to Y-Z plane at the interface between water and ribs there is practically no sound energy enters the ribs from Y-Z or X-Z planes. Rigid boundaries are assumed at the interfaces between water and ribs. When ribs are present in the acoustic pathway, most of the energy is either reflected at the water-rib interface or absorbed by the ribs so that the sound pressure immediately behind the rib is negligibly small, which was experimentally measured to be 35 dB lower than the incident pressure (Li et al. 2007).

A **weakly focused transducer** was used in the study so that the sound beam **may be** considered perpendicularly incident onto the water-rib interface along Z-direction. The boundary condition on the front face parallel to **the** X-Y plane is:

$$C_n(x, y, z_1^+) = C_n(x, y, z_1^-) \times \frac{2\rho_2 c_2}{\rho_1 c_1 + \rho_2 c_2},$$

where $\rho_1 = 993(\text{kg} / \text{m}^3)$, $c_1 = 1520(\text{m} / \text{s})$, are density and sound velocity of water, respectively; $\rho_2 = 1775 \text{kg} / \text{m}^3$, $c_2 = 3380(\text{m} / \text{s})$, are density and sound velocity of rib, respectively (Fujii et al. 1999); z_1 is distance from the transducer to the interface between water and ribs. The absorption coefficient of ribs is $\alpha_2 = 150 \text{Np} / \text{m} / \text{MHz}$ (Moros et al. 1999). Other parameters **of the ribs and the liver** are the same as those in the paper of Fujii et al. (1999).

We assume **that** the water temperature is constant and the temperature field is continuous across the rib-water interface. Using Pennes bioheat equation, we can get the temperature **distribution** inside the rib **as well as in the tissue**.

Because our experiment **was** performed on biological tissues in vitro, the heat flow from blood circulation does not exist **so that** W_b **can be negelected**, Q_m **can also be neglected in dead tissues**. Therefore, the Pennes bioheat equation can be simplified to

$$\rho_s C_s \frac{\partial \bar{T}}{\partial t} = K_s \nabla^2 \bar{T} + Q_v \quad (8)$$

where $T = T - T_0$, T_0 is the temperature before heating.

In order to calculate the temperature elevation, we **used** the alternating direction implicit (ADI) backward finite difference method to solve Eq.(8) similar to what was described by Li et al. (2007). The calculation flowchart is depicted in Fig. 1. In the i^{th} step, the calculation **was** along the X direction, line by line until the plane **was** finished, then to the $(i+1)^{th}$ step with the time being increased by Δt and the calculation **was** along the Y direction, line by line until the plane **was** finished. **Afterward**, the time was increased by Δt again and we **could start** the $(i+2)^{th}$ step, in which the calculation was along the X direction again. Similarly, in all consecutive calculations, the calculation direction **alternating** between the X and Y directions until the temperature distribution of a given time period was obtained. The scenario for the 3-D case was similar except the alternating calculation sequence was among all three X-Y-Z directions.

The numerical errors are quadratic in the space steps ΔX and ΔY , and **linear in the space step** ΔZ and time step Δt . The mesh was refined until the relative study, difference **between two consecutive** solutions was less than 0.001. In the current the step sizes in X, Y and Z axial directions **were** set to be $\Delta X = \Delta Y = 5 \times 10^{-3}$ and $\Delta Z = 5 \times 10^{-4}$, and the step size of time **was** 0.02 s.

Experimental arrangement

The experimental setup is shown in Fig 2. A programmable function generator (Agilent 33250, Loveland, CO, USA) **generated** 1.13 MHz CW signal. This signal was first amplified 55 dB by a **broadband** power amplifier (ENI A150, Rochester, NY, USA) then sent to a **weakly focused** transducer (centre frequency 1.13 MHz, diameter 35 mm, geometrical focal length 143 mm). The temperature distribution was measured by a thermocouple attached to an instrument produced by YOKOGAWA (MV100, Yokogawa Electric Co., Japan). The probe **was** mounted on a 3-D mechanical scanning system (Newport Motion Controller MM3000, Irvine, CA, USA) with a resolution of $0.1 \mu\text{m}$. Our actual scanning step was 0.1 mm. The diameter of the thermocouple is only about $50 \mu\text{m}$, so that its influence on the tissue temperature may be neglected (Eichler et al. 1978). The position uncertainty of the thermocouple was about 0.5 mm and the uncertainty of the temperature measurement was about 0.7°C .

Porcine ribs were chosen in our study because their size and acoustic properties are similar to that of human ribs (Mast et al. 1999). Two porcine ribs and a piece of liver were obtained from a local slaughterhouse. The dimensions of ribs are $7\text{mm} \times 10\text{mm} \times 40\text{mm}$, and the dimensions of the liver are $30\text{mm} \times 30\text{mm} \times 50\text{mm}$. The overlying muscles were removed by a scalpel and the ribs and liver were preserved in a 0.9% saline solution before use. Initially, the ribs were positioned 71.5 mm away from the surface of the transducer **right** in front of the liver (Fig. 2). The gap between the ribs was 10 mm. A small hole of 1 mm in diameter was made by a surgical needle along the z-direction in the liver for the thermocouple **to move** along the z-direction. The transducer, ribs and liver were immersed in distilled, degassed water at 22°C .

RESULTS

To validate the calculation procedure, we first performed calculations using parameters given by Fujii et al. (1999). As shown in Fig. 3, our prediction **matched** well with their experimental results. For the test calculation, the dimensions of the bones were: $180\text{mm} \times 50\text{mm} \times 10\text{mm}$, the diameter of the transducer **was** 10 cm, the focal length of the transducer **was** 150 mm and the center frequency of the transducer **was** 500 kHz. The **angle of incidence** of the sound wave on the bone **was** set to zero, and the heating time **was** 6 minutes.

Fig. 4 shows the theoretical **predictions** and experimental **observations** of temperature elevation versus time at the highest temperature point in the tissue for cases with and without ribs. Lines *a* and *b* represent the experimental measured and theoretically predicted temperature elevations versus time without ribs, while lines *c* and *d* are the experimental measured and

theoretically predicted temperature elevations versus time with ribs in the pathway. Relatively large difference between theory and experiments was found at the beginning for the case with ribs, but the difference becomes smaller as time goes on. After 6 minutes, the theoretical prediction was 9.26% lower than the experimentally observed values for the case without ribs, only 1.98% lower than the experimentally observed temperature elevation for the case with ribs. Such an agreement between theory and experiments is better than any published work up to date. Because we only consider hyperthermia in this study, long-time temperature distribution is more important. Therefore, our calculation procedure is quite accurate for the purpose.

One can see from the results that there was a drastic decrease in temperature when ribs blocked part of the acoustic pathway. The predicted temperature elevation **at the highest temperature point** dropped from 12.34°C down to 9.90°C while the experimental observed temperature decreased from 13.6°C down to 10.1°C. This is because the effective cross section of the beam reaching the focal point becomes smaller because the bones either reflect or absorb the acoustic energy when the beam hit them.

After 6 minutes ultrasound insonification, the simulated temperature distribution at the cross section of the highest temperature elevation point parallel to the x-y plane is shown in Fig. 5. The image on the left is for the case with ribs in the pathway, this cross section plane is 8.15 cm from the surface of the transducer. The image on the right is for the situation without ribs, and the cross section plane is 8.53 cm from the surface of the transducer. This means that the highest temperature point with ribs blocking the pathway is 0.38 cm closer to the transducer because ribs **caused** bending of the ultrasonic beam. Without ribs, the temperature distribution **has circular symmetry**, and the highest temperature elevation is 12.34 °C, while for the situation with ribs in the pathway, the temperature distribution is an ellipse and the highest temperature elevation is 9.90 °C.

The maximum calculated temperature elevation **inside ribs** is 5.83 °C, which is smaller than the maximum temperature elevation in the liver. With ribs, the temperature elevation distribution as a function of z is given Fig. 6. Along the z-line passing through the point (x,y) = (0, 0) mm, the maximum temperature elevation occurs at z = 81.5 mm, while for the z-lines passing through the point (x,y) = (6.0, 0.0) mm and (x,y) = (10.0, 0.0) mm, the maximum temperature elevations both occur at z = 72.3 mm. As a comparison, the temperature elevation without ribs along these three z-lines are given in Fig.7.

The influence of rib's position on the location of maximum temperature after 80 seconds insonification is shown in Table. I. When the distance between the transducer and ribs changes from 3.7 cm to 9.45 cm, the position of maximum temperature changes from 7.86 cm to 9.68 cm **from the transducer surface**, and the temperature elevation increases from 3.47°C to 16.91 °C. Notice that when the distance between the transducer and the ribs increased 5.75 cm, the position of maximum temperature only increased by 1.82 cm, but the temperature elevation increased by 13.44 °C. This is because the sound wave is transmitted through the gap between the two ribs. When the distance between the transducer and ribs becomes smaller, more sound energy is blocked by the ribs because the transducer size is bigger than the gap. As the ribs moves away from the transducer, the focused beam becomes narrower so that more acoustic energy can pass through the gap. At the same time, because the sound wave propagation distance in the lossy biological tissue also increases when the ribs (on the liver surface) are closer to the transducer surface, the temperature elevation becomes lower and it also takes longer time for the temperature to increase. Lines a-e in Fig. 7 represent cases for transducer-rib distance of 9.45, 8.3, 6.0, 4.85 and 3.7 cm, respectively, and Fig. 8 shows the temperature increase with time for different transducer-rib distances.

One can see from Table I that the maximum temperature elevation in ribs also increased with the distance between the transducer and ribs. Table II shows the influence of ribs on the maximum temperature elevations in the liver and in ribs after 80 second insonification.

If we want the temperature maximum to appear at a fixed distance of 4.16 cm deep in the liver, the transducer must be put at 4.59 cm and 3.7 cm away from the liver surface for cases with and without ribs, respectively. That is to say, if we want to treat the liver cancer at fixed point, we must reduce the distance between the transducer and the tumor when the ribs are present. Our results showed that the maximum temperature elevation changed from 5.64 °C to 3.47 °C due to the presence of ribs.

The influence of **the** rib width (0.4 cm-1.0 cm for normal people, Mohr et al. 2007) on the position **and magnitude** of maximum temperature after 80 seconds insonification are given in Table III, for which the distance between the transducer and ribs is 7.5 cm and the gap between ribs is 1.2 cm. From Table III the influence of rib width on the position **and magnitude** of maximum temperature is not significant. This is because the temperature elevation is mainly caused by the sound energy transmitted through the gap between ribs while the energy diffracted from the edge of the ribs is **negligible**.

For humans, the gap between ribs is in the range of 0.6 - 1.5 cm (Nunn et al. 1980) and it increases with age before the age of 25. Research data showed that the percentage of people with narrow gap of less than 1 cm is 93.4% (Liu 2000), which makes ultrasound treatment troublesome. Therefore, understanding the influence of rib gap on the temperature elevation is very useful. As shown in Table IV, when the gap between ribs increased from 0.6 cm to 1.5 cm, the position of maximum temperature moved away from the transducer from 7.74 cm to 9.02 cm and the highest temperature elevation also increased from 7.57 °C to 13.56 °C. For wider gap between ribs, more ultrasonic power can enter the tissue, so that higher temperature can be produced in the liver. Fig. 9 shows the highest temperature elevation with time for the gap size from 0.6 cm to 1.5 cm. When the gap size is 0.6 cm, the highest temperature elevation is 7.57°C after 80 second treatment, while for a 1.5 cm gap, the maximum temperature rise becomes 13.56 °C for the same time period of treatment.

Table V shows the effect of transducer f-number on the position and peak value of the maximum temperature. **One can see that** with the decrease of the transducer f-number, both the maximum temperature and its position increase.

DISCUSSIONS AND SUMMARY

In summary, **the** temperature distributions in tissues and bones were calculated by the Pennes bioheat equation. We showed that the presence of ribs greatly affect the temperature distribution, both in terms of peak value and position. Our theoretical predictions agree well with our experimental measurements using a weakly focused ultrasonic transducer.

We found that the temperature distribution and the position of maximum temperature are strongly influenced by the distance between the transducer and the ribs, while the gap between ribs is a determining factor for the highest achievable temperature. However, the width of the rib has little effect on the position and peak value of the maximum temperature.

The maximum temperature elevation inside ribs **also** increases with the distance between the transducer and ribs but decreases with the increase of the rib gap size as well as the transducer f-number.

Our results **indicated** that the most effective way to control the temperature in ultrasound induced hyperthermia is to use appropriate f-number focused transducer and adjust the distance

between the transducer and ribs using **water as a medium in between**. Based on the width of rib and the rib gap size of a particular patient, doctors should determine the sound intensity of the transducer and the distance between the transducer and the ribs. Our theoretical treatment presented here may provide a useful tool to help doctors making **such** decisions.

Acknowledgments

Financial support for this work was provided by the NIH under grant No. P41-EB21820, National Natural Science Foundation of China under grant No.10674066 and State Key Laboratory of Acoustics, Chinese Academy of Sciences.

REFERENCES

- Aubry JF, Pernot M, Marquet F, Tanter M, Fink M. Transcostal high intensity focused ultrasound: ex vivo adaptative focusing feasibility study. *Phys. Med. Biol* 2008;53(11):2937–2951. [PubMed: 18475006]
- Botros YY, Ebbini ES, Volakis JL. Two-Step Hybrid Virtual Array-Ray (VAR) Technique for Focusing through the Rib Cage. *IEEE Trans Ultrason Ferroelect and Freq Contr* 1998;45(4):989–1000.
- Botros, YY.; Volakis, JL.; Ebbini, ES. Analysis and synthesis of multiple-focus phased array heating patterns through the rib cage: a simulation model; Proceedings of the thirteenth National Radio Science Conference; 1996;
- Chaussy C, Thüroff S, Rebillard X, Gelet A. Technology insight: High-intensity focused ultrasound for urologic cancers. *Nat Clin Pract Urol* 2005;2(4):191–198. [PubMed: 16474762]
- Chavrier F, Chapelon JY, Gelet A, Cathignol D. Modeling of high-intensity focused ultrasound-induced lesions in the presence of cavitation bubbles. *J. Acoust Soc. Am* 2000;108(1):432–440. [PubMed: 10923905]
- Diederich CJ, Hynynen K. Ultrasound technology for hyperthermia. *Ultrasound Med Biol* 1999;25(6):871–887. [PubMed: 10461714]
- Eichler J, Knof J, Lenz H, Salk J, Schafer G. Temperature distribution in tissue during laser irradiation. *Rad. And Environm Biophy* 1978;15:277–287.
- Fujii M, Sakamoto K, Toda Y, Negishi A, Kanai H. Study of the cause of the temperature rise at the muscle-bone interface during ultrasound hyperthermal. *IEEE Trans Biomed Eng* 1999;46(5):494–504. [PubMed: 10230128]
- Goss SA, Johnston RL, Dunn F. Comprehensive compilation of empirical ultrasonic properties of mammalian tissues. *J Acoust Soc. Am* 1978;64(2):423–457. [PubMed: 361793]
- Gunnar A. Current Status of HIFU and Cryotherapy in Prostate Cancer – A Review. *European urology* 2006;50:927–934. [PubMed: 16971038]
- Hill CR. Calibration of ultrasonic beams for biomedical applications. *Phys Med. Biol* 1970;15(2):241–248. [PubMed: 5484867]
- Illing RO, Kennedy JE, Wu F, Ter Haar GR, Protheroe AS, Friend PJ, Gleeson FV, Cranston DW, Phillips RR, Middleton MR. The safety and feasibility of extracorporeal high-intensity focused ultrasound (HIFU) for the treatment of liver and kidney tumors in a western population. *Br J Cancer* 2005;93(8):890–895. [PubMed: 16189519]
- Kuznetsov VP. Equation of nonlinear acoustics. *Sov Phys Acoust* 1970;16(4):467–470.
- Lapidus, L.; Pinder, GF. Numerical solution of partial differential equations in science and engineering. Wiley; New York: 1982.
- Li CX, Xu GL, Jiang ZY, Li JJ, Luo GY, Shan HB, Zhang R, Li Y. Analysis of clinical effect of high-intensity focused ultrasound on liver cancer. *World J Gastroenterol* 2004;10(15):2201–2204. [PubMed: 15259065]
- Li JL, Liu XZ, Zhang D, Gong XF. Influence of ribs on the nonlinear sound field of therapeutic ultrasound. *Ultrasound Med Biol* 2007;33(9):1413–1420. [PubMed: 17630093]
- Liu HL, Chang H, Chen WS, Shih TC, Hsiao JK, Lin WL. Feasibility of transrib focused ultrasound thermal ablation for liver tumors using a spherically curved 2D array: A numerical study. *Med Phys* 2007;34(9):3436–3448. [PubMed: 17926945]

- Liu Z. Re-evaluation of normal people' rib cleft (801 X-ray measurement of normal people' rib cleft). *Chinese Journal of medical imaging technology* 2000;16(7):589.
- Mast TD, Hinkelman LM, Metlay LA, Orr MJ, Waag RC. Simulation of ultrasonic pulse propagation, distortion, and attenuation in the human chest wall. *J Acoust Soc Am* 1999;106(6):3665–3677. [PubMed: 10615705]
- Matsumine A, Kusuzaki K, Matsubara T, Shintani K, Satonaka H, Wakabayashi T, Miyazaki S, Morita K, Takegami K, Uchida A. Novel hyperthermia for metastatic bone tumors with magnetic materials by generating an alternating electromagnetic field. *Clin Exp Metastasis* 2007;24(3):191–200. [PubMed: 17487560]
- Mohr M, Abrams E, Engel C, Long WB, Bottlang M. Geometry of human ribs pertinent to orthopedic chest-wall reconstruction. *Journal of Biomechanics* 2007;40:1310–1317. [PubMed: 16831441]
- Moros EG, Fan XB, Straube WL. Ultrasound power deposition model for the chest wall. *Ultrasound Med Biol* 1999;25(8):1275–1287. [PubMed: 10576270]
- Moros EG, Novak P, Straube WL, Kolluri P, Yablonskiy DA, Myerson RJ. Thermal contribution of compact bone to intervening tissue-like media exposed to planar ultrasound. *Phys Med Biol* 2004;49(6):869–886. [PubMed: 15104313]
- Nunn JF, Slavin G. Posterior intercostals nerve block for pain relief after cholecystectomy. Anatomical basis and efficacy. *Br J Anaesth* 1980;52:253–260. [PubMed: 7370141]
- Pennes HH. Analysis of tissue and arterial temperatures in the resting human forearm. *J Appl Physiol* 1948;1(2):93–122. [PubMed: 18887578]
- Qian SY, Wang HZ. Theoretical study of the thermal effect in biological medium generated by focused ultrasound source. *Acta Physica Sinica* 2001;50(3):501–506.
- Tanter M, Thomas JL, Fink M. Focusing and steering through absorbing and aberrating layers: Application to ultrasonic propagation through the skull. *J Acoust Soc Am* 1998;103(5):2403–2410. [PubMed: 9604342]
- ter Haar GR. Turning up the Power: High Intensity Focused Ultrasound (HIFU) for the Treatment of Cancer. *Ultrasound* 2007;15(2):73–77.
- Wu F, Wang ZB, Chen WZ, Zhu H, Bai J, Zou JZ, Li KQ, Jin CB, Xie FL, Su HB. Extracorporeal high intensity focused ultrasound ablation in the treatment of patients with large hepatocellular carcinoma. *Ann Surg Oncol* 2004;11(12):1061–1069. [PubMed: 15545506]
- Wu F, Wang ZB, Chen WZ, Zou JZ, Bai J, Zhu H, Li KQ, Xie FL, Jin CB, Su HB, Gao GW. Extracorporeal focused ultrasound surgery for treatment of human solid carcinomas: early Chinese clinical experience. *Ultrasound Med Biol* 2004;30(2):245–260. [PubMed: 14998677]
- Wu JR, Du GH. Temperature elevation generated by a focused Gaussian ultrasonic beam at a tissue-bone interface. *J Acoust Soc Am* 1990;87(6):2748–2755. [PubMed: 2197307]

Fig. 1.
Alternating direction implicit (ADI) backward finite difference method.

Fig. 2.
Experimental setup.

Fig. 3. Comparison of temperature elevation distribution along z-direction using parameters in the paper of Fujii et al. (1999). Line: calculation results using our model; • experimental results from the reference.

Fig. 4.

Theoretically predicted and experimentally measured temperature elevations as a function of time. a). experimental results without rib; b). theoretical results without ribs; c). experimental results with ribs; d). theoretical results with ribs. Parameters used in the simulation are as follows: rib width=1.0 cm, gap between ribs=1.0 cm, distance between the transducer and the ribs=7.15 cm.

Fig. 5. Temperature distribution at the cross section including the highest temperature elevation point. Position of maximum temperature is 8.15 cm from the transducer with ribs and 8.53 cm without ribs.

Fig. 6. Temperature elevation distribution along z-direction with ribs: a. along the line passing through $(x, y) = (0.0, 0.0)$ mm; b. along the line passing through $(x, y) = (6.0, 0.0)$ mm; c. along the line passing through $(x, y) = (10.0, 0.0)$ mm.

Fig. 7. Temperature elevation distribution along z-direction without ribs: a. along the line passing through $(x, y) = (0.0, 0.0)$ mm; b. along the line passing through $(x, y) = (6.0, 0.0)$ mm; c. along the line passing through $(x, y) = (10.0, 0.0)$ mm.

Fig. 8.

Maximum temperature elevation with time at fixed rib width (=1.0 cm) and gap between ribs (=1.2 cm) for cases with ribs being put at different positions from the transducer (a. 9.45 cm; b. 8.3 cm; c. 6cm; d. 4.85 cm; e. 3.7 cm). The corresponding positions of maximum temperature are: a. 9.68 cm; b. 8.90 cm; c. 8.19 cm; d. 7.95 cm; and e. 7.86 cm.

Fig. 9.

Maximum temperature elevation with time for different size gaps between ribs (a. 1.5 cm; b. 1.2 cm; c. 1.0 cm; d. 0.85 cm; e. 0.7 cm; f. 0.6 cm). The corresponding positions of maximum temperature are: a. 9.02 cm; b. 8.70 cm; c. 8.62 cm; d. 8.39 cm; e. 8.04 cm; and f. 7.74 cm. Here rib width=1.0 cm, distance between the transducer and ribs=7.5 cm.

Table. I

Influence of the distance between the transducer and ribs on the position and amplitude of the maximum temperature elevation after 80 second insonification (gap between the ribs=1.2 cm, rib width=1.0 cm)

Distance between transducer and ribs (cm)	Position of maximum temperature from transducer(cm)	Maximum temperature elevation in liver (°C)	Maximum temperature elevation in ribs (°C)
3.7	7.86	3.47	3.52
4.85	7.95	4.97	4.27
6.0	8.19	7.78	5.14
8.3	8.90	14.20	7.31
9.45	9.68	16.91	8.20

Table. II

Influence of ribs on the maximum temperature elevations after 80 second insonification when the distances between the water/liver interface and the position of maximum temperature are fixed at different places (gap between ribs=1.2 cm, rib width=1.0 cm)

	Distance between transducer and the point of maximum temperature (cm)	Distance between water/liver interface and position of maximum temperature (cm)	Maximum temperature elevation in liver (°C)	Maximum temperature elevation in ribs (°C)
No rib	8.75	4.16	5.64	
With rib	7.86	4.16	3.47	3.52
No rib	8.80	3.1	7.09	
With rib	7.95	3.1	4.97	4.27
No rib	8.92	2.19	9.83	
With rib	8.19	2.19	7.78	5.14

Table. III

Influence of **rib** width on the position and amplitude of the maximum temperature elevation after 80 second insonification. Distance between the transducer and ribs=7.5 cm, gap between the ribs=1.2 cm

Rib width (cm)	Position of maximum temperature (cm)	Maximum temperature elevation in liver (°C)	Maximum temperature elevation in ribs (°C)
0.4	8.73	12.81	6.32
0.6	8.72	12.72	6.35
0.8	8.71	12.71	6.41
1.0	8.71	12.71	6.43

Table. IV

Influence of gap size between ribs on the position and amplitude of the maximum temperature **elevation** after 80 second insonification. Distance between transducer and ribs=7.5 cm, rib width=1.0 cm

Gap between ribs (cm)	Position of maximum temperature (cm)	Maximum temperature elevation in liver (°C)	Maximum temperature elevation in ribs (°C)
0.6	7.74	7.57	10.24
0.7	8.04	8.64	8.94
0.85	8.39	10.15	8.73
1.0	8.62	11.05	7.41
1.2	8.71	12.25	6.43
1.5	9.02	13.56	5.27

Table. V

Influence of the transducer f-number on the position and amplitude of the maximum temperature after 80 second insonification. Distance between transducer and ribs=7.5 cm, gap between ribs=1.2 cm, rib width=1.0 cm. The geometrical focal length of the transducer is 143 mm, and the diameter of the transducer is 30 cm, 35 cm, 40 cm and 50 cm, respectively

f-number	Position of maximum temperature (cm)	Maximum temperature elevation in liver (°C)	Maximum temperature elevation in ribs (°C)
4.8	7.62	11.02	5.01
4.1	8.71	12.25	6.43
3.6	8.98	13.27	7.21
2.9	9.20	14.51	8.32