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Simulations of adaptive temperature control with self-focused hyperthermia system for tumor treatment

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ABSTRACT

The control problem in ultrasound therapy is to destroy the tumor tissue while not harming the intervening healthy tissue with a desired temperature elevation. The objective of this research is to present a robust and feasible method to control the temperature distribution and the temperature elevation in treatment region within the prescribed time, which can improve the curative effect and decrease the treatment time for heating large tumor ($\geqslant$2.0 cm in diameter). An adaptive self-tuning-regulator (STR) controller has been introduced into this control method by adding a time factor with a recursive algorithm, and the speed of sound and absorption coefficient of the medium is considered as a function of temperature during heating. The presented control method is tested for a self-focused concave spherical transducer (0.5 MHz, 9 cm aperture, 8.0 cm focal length) through numerical simulations with three control temperatures of 43 °C, 50 °C and 55 °C. The results suggest that this control system has adaptive ability for variable parameters and has a rapid response to the temperature and acoustic power output in the prescribed time for the hyperthermia interest. There is no overshoot during temperature elevation and no oscillation after reaching the desired temperatures. It is found that the same results can be obtained for different frequencies and temperature elevations. This method can obtain an ellipsoid-shaped ablation region, which is meaningful for the treatment of large tumor.

1. Introduction

In high-intensity focused ultrasound (HIFU) surgery, the heating effect of ultrasound is used to cause lesions and coagulate the cancerous tissue, and the temperature of the cancerous tissue is raised up to 50–90 °C for several seconds [1,2]. One important issue is that the heat produced by HIFU should be controlled selectively to obtain a desired temperature distribution in the treatment area, and the temperature in healthy tissue should stay within 43 °C. A variety of control and optimization methods has been proposed to control the temperature in tissue. These methods include for example standard proportional–integral–derivation (PID) [3], linear quadratic regulator (LQR) [4], inverse dynamics [5], fuzzy logic [6] and model predictive control (MPC) [7] approaches. In addition, an adaptive self-tuning-regulator (STR) controller has attracted much attention and is used for ultrasound hyperthermia system [8–10]. The STR is an adaptive controller, which can specify the structure of model and corresponding parameters with a recursive least square estimation (RLSE) method and automatically converge the actual output to the desired temperature. To ensure reliable temperature control, another important issue for HIFU surgery is the understanding the change of acoustical properties at higher temperatures (50–100 °C). Tissue properties such as sound speed and absorption coefficient would change with temperature during the course of a single focused ultrasound treatment [11,12]. For example, absorption coefficient increased at temperatures higher than 50 °C, The rate of change of tissue attenuation as a function of temperature was between 0.239 and 0.291 Np/(m MHz °C) for muscle, liver, and kidney, respectively [11]. However, there is little documentation concerning the effect of change of tissue properties on temperature control.

Our earlier works demonstrated the feasibility of a stationary temperature profile obtained with the STR [10]. In the present paper, we extend these studies by adding a time factor to the STR controller with a recursive algorithm. This goal is to control the temperature elevation in specified time during HIFU heating. Thus it can give a reasonable time frame for heating large tumor ($\geqslant$2 cm in diameter). Simultaneously tissue properties such as absorption coefficient and speed of sound are taken into account and are as a function of temperature during heating with a self-focused concave spherical transducer. The control system performance is evaluated with 1D simulation.

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In this article, we firstly describe the spheroidal beam equation (SBE) and the finite Fourier integral transformation to the bioheat equation and control method. Then, the simulations of temperature control and acoustic output power response are performed using the theory and method introduced. We give a discussion and summarize our studies finally.

2. Theory and methods

2.1. Spheroidal beam equation (SBE)

Almost all the work begins with the Khokhlov–Zabolotskaya–Kuznetsov (KZK) equation [13,14] that is extensively used as a model equation for describing the combined effects of diffraction, absorption, and nonlinearity in directional beams. The KZK equation is derived under the paraxial approximation. Therefore, its applicability to focused beams is limited to small opening angles of aperture. Another paraxial approximation algorithm is time-averaged wave envelopes (TAWEs) method [15] which can predict nonlinear beam propagation based on the solution of the second order nonlinear differential wave equation [16,17] for lossy media, arbitrarily shaped plane and focused sources. But its computation time depends on several parameters, including the source geometry, dimensions, fundamental resonance frequency, excitation level as well as the strength of the medium nonlinearity. The spheroidal beam equation (SBE) [18] is a parabolic equation using an oblate spheroidal coordinate system [19] and has a specific application to a theoretical prediction on focused, high-frequency beams from a circular aperture, whose upper limit of the applicability is at least 40° for the half-aperture angle. It is feasible through a finite difference scheme and takes no more than 10 min of run time to obtain all harmonic components using a standard PC. Fig. 1 shows the oblate spheroidal coordinate system, in which the variables (σ, η, φ) are related to the rectangular coordinates (x, y, z) by

$$\begin{cases} x = b\sqrt{(1+\sigma^2)(1-\eta^2)}\cos\varphi \\ y = b\sqrt{(1+\sigma^2)(1-\eta^2)}\sin\varphi \\ z = b\sigma\eta \end{cases} \tag{1}$$

with $-\infty < \sigma < \infty$, $0 \leqslant \eta \leqslant 1$, $0 \leqslant \varphi < 2\pi$, where 2 b is the interfocal length. The focused field is usually divide into two regions, i.e. $\sigma < \sigma_0 < 0$ and $\sigma_0 \leqslant \sigma$, where $\sigma = \sigma_0$ (=const < 0) denotes a specific transition location. The former region is close to the source where spherically converging waves are predominant and the spherical coordinates are preferable to the rectangular ones for field analysis; the latter region is near the focus where the paraxial or planar parabolic approximation is useful. For each region, different retarded time is introduced to observe progressive waves in a frame moving with speed c. In terms of σ, η, and θ, the Westervelt equation [20]

$$\nabla^2 p - \frac{1}{c}\frac{\partial^2 p}{\partial t^2} + \frac{2\alpha}{c\omega^2}\frac{\partial^3 p}{\partial t^3} + \frac{\beta}{\rho_0 c^4}\frac{\partial^2 p^2}{\partial t^2} = 0 \tag{2}$$

can be reduced to the nonlinear SBE and is described as below [18]

$$\begin{aligned} &\frac{\partial^2 \bar{p}}{\partial\tau_s\partial\sigma} + \frac{1}{2}\frac{\sin 2\theta}{\sigma(1+\sigma^2)}\frac{\partial^2 \bar{p}}{\partial\tau_s\partial\theta} + \frac{\varepsilon\sqrt{\sigma^2-\sin^2\theta}}{\sigma(1+\sigma^2)} \\ &\times\left(\frac{\partial^2 \bar{p}}{\partial\theta^2} + \cot\theta\frac{\partial\bar{p}}{\partial\theta}\right) + \frac{E}{\sigma}\frac{\partial\bar{p}}{\partial\tau_s} \\ &= \frac{-\sqrt{\sigma^2+\sin^2\theta}}{\sigma}\times\left(\alpha b\frac{\partial^3\bar{p}}{\partial\tau_s^3} + \frac{b}{2l_D}\frac{\partial^2\bar{p}^2}{\partial\tau_s^2}\right)E \qquad (\sigma < \sigma_0 < 0) \end{aligned} \tag{3}$$

$$\begin{aligned} &\frac{\partial^2 \bar{p}}{\partial\tau_p\partial\sigma} - \frac{\sigma}{1+\sigma^2}\sin\theta\frac{\partial^2 \bar{p}}{\partial\tau_p\partial\theta} - \frac{\varepsilon(2-\cos\theta)}{1+\sigma^2}\times\left(\frac{\partial^2 \bar{p}}{\partial\theta^2} + \cot\theta\frac{\partial\bar{p}}{\partial\theta}\right) \\ &= \left(\alpha b\frac{\partial^3\bar{p}}{\partial\tau_p^3} + \frac{b}{2l_D}\frac{\partial^2\bar{p}^2}{\partial\tau_p^2}\right)E \quad (\sigma \geq \sigma_0, \sigma_0 < 0) \end{aligned} \tag{4}$$

Here σ_0 is a specific transition location where we can transform from the spherical to plane wave region, and

$$\begin{cases} \bar{p} = \frac{p}{p_0}, \quad \tau_p = t - \frac{b\sigma\eta}{c}, \quad \varepsilon = \frac{1}{2kb}, \quad \eta = \cos\theta \\ \tau_s = \tau_p + kb\left(\sqrt{\sigma_0^2 + \sin^2\theta} + \sigma_0\cos\theta\right) \\ E = \frac{\sigma^2+\cos^2\theta}{1+\sigma^2}, \quad \alpha = \frac{\delta\omega^2}{2c^3}, \quad l_D = \frac{\rho_0 c^3}{\beta p_0 \omega} \end{cases} \tag{5}$$

where c is sound speed, ρ_0 is the ambient density of medium, ω is the angular frequency of ultrasound, δ is the sound diffusivity, α is the attenuation, β is the nonlinear coefficient of medium, p_0 (1.013 e5 Pa) and p are the amplitude of sound pressure on the source face and in the medium, respectively. Parameters α and c for all calculations are the variables relating to the temperature.

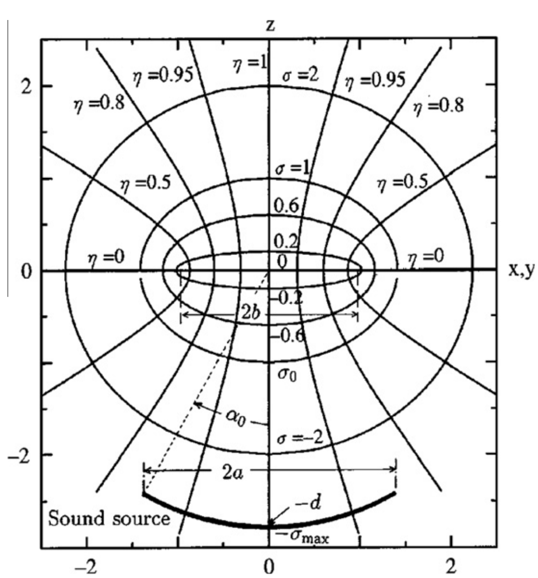

Fig. 1. Oblate spheroidal coordinate system.

2.2. The finite Fourier integral transform to bioheat equation

Once the pressure is determined with Eqs. (3) and (4), the heat transfer in tissue from the ultrasound field is determined by applying Pennes' bioheat transfer equation (BHTE) [21]:

$$\rho_t C_t \frac{\partial T}{\partial t} = K_t \nabla^2 T - W_b C_b (T - T_\infty) + Q \tag{6}$$

where T is the tissue temperature; ρ_t, C_t, and K_t are the density, heat capacity, and thermal conductivity with the subscripts t and b referring, respectively, to tissue and blood domain; T_∞ refers to the temperature at large distance from the acoustically induced thermal lesion, which corresponds to the initial condition value of 37 °C. The first term on the right-hand side of (6) accounts for heat diffusion; the second term of (6) is responsible for blood perfusion losses, with W_b being the perfusion rate. The term Q is the rate of heat production per unit volume due to the ultrasonic field. The three dimensional representation of this equation in Cartesian coordinates is:

$$\begin{aligned} \rho_t C_t \frac{\partial T}{\partial t} = K_t\left(\frac{\partial^2 T}{\partial x^2} + \frac{\partial^2 T}{\partial y^2} + \frac{\partial^2 T}{\partial z^2}\right) - W_b C_b (T - T_\infty) \\ + Q(x, y, z; t) \end{aligned} \tag{7}$$