Hyperthermia Treatment and Temperature Control of the Breast Cancer Therapy Applying Ultrasound Induced Heating

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Abstract

High intensity ultrasound has great potential in non-invasive treatment of deep-seated tumors. In a hyperthermia treatment the main goal is to elevate tumor temperature to a cytotoxic level without overheating the intervening healthy tissues. To produce deep-seated lesions using ultrasound requires high power and intensity levels at focal zone. Conventionally, appropriate transducers are used which require suitable apertures for ultrasound hyperthermia. Such transducers result in strong off-focus maxima, where the resulting hot spots can destroy tumors.
This paper describes a new approach in transducer design and acoustic field calculation and consequently temperature tracking during a hyperthermia treatment of breast tumors. For this purpose the linear acoustic pressure equation is used to calculate the transmitted energy in the focal zone. As the vibration of tissue causes the increase of the temperature inside the tissue and destruction of tumor happens, Pennes' Bio-heat Transfer equation is used for temperature estimation during the treatment. The numerical Finite Element Method is applied to do a precise mathematical calculation.
The results of calculations and also simulation outcomes have been mentioned in the text.

Keywords: *Focused Ultrasound, FEM, Acoustic Pressure, BHTE, Breast Cancer.*

Introduction

High intensity focused ultrasound direction to a medium causes the absorption of energy and turning it to heat. The main goal is to direct ultrasound to tumors and heat cancerous tissues. The ultrasound transducer is located outside the body hence the treatment is bloodless surgery and non-invasive.
The desired temperature in tumor is often 50-60 ºC. Although lower temperatures could also be used, the use of high temperatures can reduce the treatment time significantly [1,2,3,4]. The temperature distribution optimizations in wave field induced heating problems in medical applications are usually done by optimizing the specific absorption rate (SAR) or using PID type controllers with pre-focused ultrasound fields.
While some cases like breast tumors can be treated with relatively simple computational models, the treatment of brain tumors poses significant problems. This is due to the geometrical problems and the high amount of attenuation in the skull.

The Mathematical Model

Acoustic Field Calculation

Linear acoustic wave propagation and scattering in quiescent heterogeneous media is characterized by the following wave equation:

$$\nabla^2 p - \frac{1}{c^2}\frac{\partial^2 p}{\partial t^2} = 0 \qquad (1)$$

which is arising from the equations of hydrodynamics by assuming an incompressible fluid. The initial and boundary conditions are as follows:

$$\varphi(x,0) = \varphi_0 \quad , \quad x \in \Omega \qquad (2)$$

$$\frac{\partial \varphi}{\partial t}(x,0) = 0 \quad , \quad x \in \Omega \qquad (3)$$

$$\frac{\partial p}{\partial n} = -\frac{1}{c}\frac{\partial p}{\partial t} \quad , \quad x \in \partial\Omega_{T^c} \qquad (4)$$

where a simpler relation between φ and p is used:

$$p - p_0 = \rho_0 \frac{\partial \varphi}{\partial t} \qquad (5)$$

The boundary $\partial\Omega$ consists of two disjointed parts: $\partial\Omega_T$ and $\partial\Omega_{T^c}$, where $\partial\Omega_T$ is the part of the boundary on which the ultrasound transducer is located. On $\partial\Omega_T$ the pressure wave we want to study is implemented. And $\partial\Omega_{T^c}$ is the non-reflecting boundary. Using centered differences for the time derivatives, we get the semi-discrete scheme:

$$\nabla^2 p^n - \frac{1}{c^2\Delta t^2}(p^{n+1} - 2p^n + p^{n-1}) = 0, \quad x \in \Omega \qquad (6)$$

The Galerkin finite element method will yield a set of *m* equations for *m* unknowns p_i^{n+1}:

$$\sum_{i=1}^{m}\left[\frac{1}{c\Delta t}\int_{\partial\Omega_{T^c}} N_i N_j d\Gamma + \frac{1}{c^2\Delta t^2} N_i N_j\right] p_i^{n+1} = \frac{1}{c\Delta t}\int_{\partial\Omega_{T^c}} p_n N_j d\Gamma - (\nabla p^n, \nabla N_j) + \frac{1}{c^2\Delta t^2}(2p^n - p^{n-1}, N_j) \quad , \quad j = 1 \ldots m \qquad (7)$$

which can be solved explicitly at each time level.

Temperature Control

A thermal model of the tumor and intervening tissue is obtained by using the finite element method (FEM) to approximate the Pennes' BHTE [5], in which the temperature, *T*, at any location in a volume of interest satisfies the partial differential equation:

$$\rho c \frac{\partial T}{\partial t} = \nabla . k \nabla T - w_b c_b (T - T_a) + Q \qquad (8)$$

where ρ is the tissue density, c is the tissue specific heat capacity, c_b is the blood specific heat capacity, T is temperature, T_a is the arterial blood temperature, t is time, k is the thermal conductivity, w_b is a volumetric blood perfusion rate, and Q is the power density delivered to tissue by external means. The heat source term for the time-harmonic acoustic pressure is [6]:

$$Q = \frac{\alpha |p|^2}{\rho c} \qquad (9)$$

The Bio-Heat transfer equation is discretized according to the usual semi-discrete scheme in which the spatial variable is handled with the Galerkin scheme and the resulting system of ordinary differential equations with appropriate (implicit) schemes such as backward Euler [7]. The semi-discrete FEM approximation for the Bio-Heat equation can be written in the form:

$$T^{(e)}(x, y, z, t) = \sum_{i=1}^{n} N_i(x, y, z) T_i(t) \qquad (10)$$

$$M\left\{\dot{T}\right\} + [K]\{T\} + w_b c_b (T - T_a) = \{Q\} \qquad (11)$$

where *M* is the mass matrix, $\dot{T} = \frac{dT}{dt}$, and *K* is the stiffness matrix.

Results

For the purpose of temperature rise and high amount of energy absorption, frequency is a dominant point. Using a range of frequency below 1 MHz mostly the frequencies between 400 KHz-600 KHz, makes the temperature rise up to 60 ºC, which is desired in hyperthermia treatment

The results out of simulation of temperature distribution in the focal zone for the purpose of tumor destruction show that the chosen frequency ($\nu = 410$ KHz)-for obtaining the desired intensity delivery- satisfies the expected temperature value.

Due to Fig. 1 it is considered that temperature changes from 37 ºC in boundary to 115.405 ºC in the focal point that is the location of tumor in the breast. Spatial distribution of temperature implies that the energy transmission to the focal region provides the desired temperature in a condition that surrounding healthy tissues stay safe and undamaged.

Fig. 2 indicates the relation of temperature increase period with time. Due to this diagram the temperature rises to 60 ºC in a period of 30 seconds after radiation is finished. So in 30 seconds we have been able to increase the temperature from 37 ºC in boundaries to 60 ºC in the target. Fig. 3 shows the temperature changes from the focal zone to the boundaries. Due to this figure it is understood that in the focal point temperature is 115.45 ºC and as we move toward the boundaries temperature decreases to 37 ºC, which is normal temperature of the healthy tissues. Also it should be noticed that the blood perfusion rate of the breast tissue has been considered in the simulation of the treatment temperature control. The characteristics of the designed temperature are as table 1. As the main purpose of this study is temperature control of breast cancer therapy, the details of transducer designation

and the related calculations have not been mentioned.

Conclusion

In this paper a brief study of ultrasound therapy, mathematical models for ultrasound therapy, and at last results out of simulations have been presented to destroy the tumors of breast a critical cancer prevalent among women. The presented results were tested with numerical simulations in 3D. The mathematical models which are used for ultrasound therapy modeling discusses wave field and thermal therapy modeling. The wave field is modeled using acoustic pressure equation using FEM, and due to acoustic pressure calculation, the acoustic intensity transmitted to the focal region is computed. Thermal treatment process modeling has been done using Pennes' bioheat transfer equation. Temperature distribution and the process of temperature rise in the focal region in relation with time and distance is presented.

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Number of elements	192
Fixed focus	45 mm
Radius of curvature	30 mm
Frequency	410 KHz

TABLE 1. Designed transducer characteristics

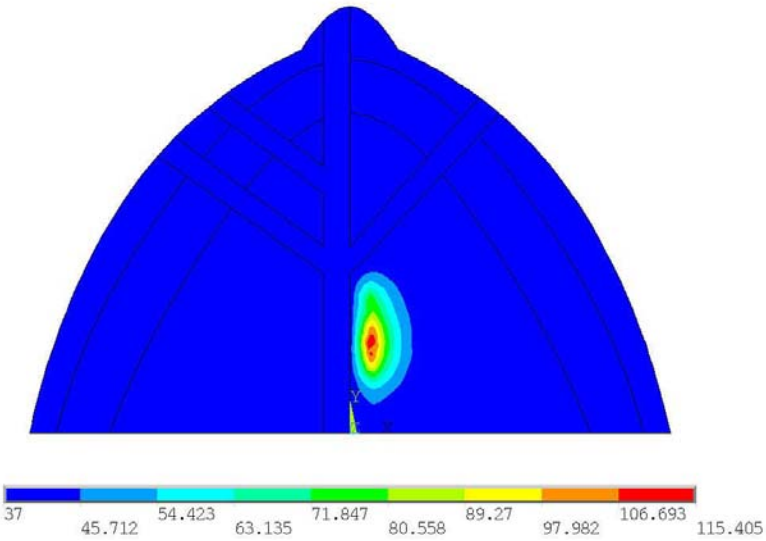

Figure 1: Spatial temperature distribution

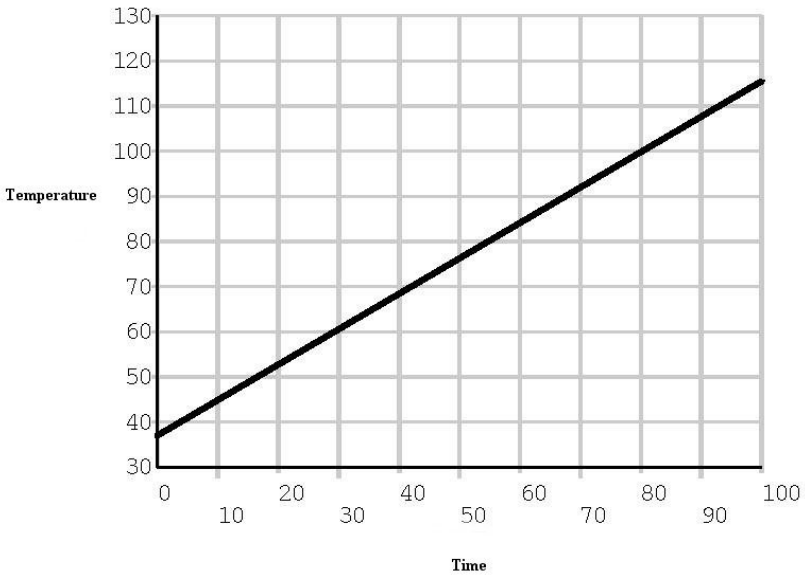

Figure 2: Temperature changes due to time

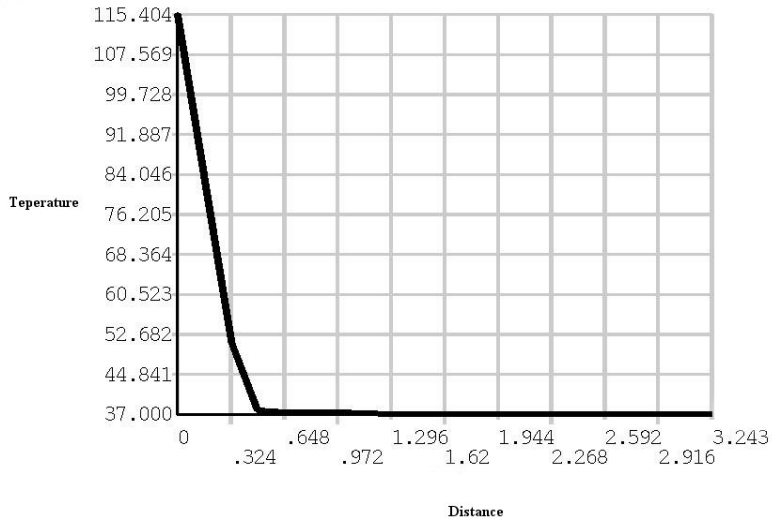

Figure 3: Temperature distribution from the focal point to the surrounding tissues