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Analysis of the thermal response and requirement for power dissipation in magnetic hyperthermia with the effect of blood temperature

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ABSTRACT

Magnetic hyperthermia has been an oncological technique. To know the heat transport occurring in biological tissues during hyperthermia treatment is essential for its improvement. The biological tissue can be treated as a fluid saturated porous medium. Therefore, this work applies the generalized dual-phase-lag model of bioheat transfer, which was derived from a two-temperature model, to describe the behavior of bioheat transfer in the tumor and its surrounding healthy tissue with transient blood temperature during hyperthermia treatment. A hybrid numerical scheme based on the Laplace transform is proposed to solve the present problem. The effect of porosity, coupling factor, metabolic heat generation, time-dependent blood temperature, and size of tumor on the thermal response is investigated. The suitable power dissipation is also estimated for a sustained temperature which could ablate the tumor without damaging surrounding healthy tissues. In order to show the effect of blood temperature, the comparison between the results with the transient blood temperature and those with the constant blood temperature is done.

1. Introduction

The principle of medical treatment of tumors is to eliminate only tumor cells from normal cells. However, it is difficult to distinguish them in surgical procedure. On tumor-selective targeting, hyperthermia should be superior to other therapeutic techniques [1]. At the same time, magnetic fluid hyperthermia had the maximum potential for such selective targeting [2]. In magnetic tumor hyperthermia, fine magnetic particles are localized at the tumor tissue. Then, an alternating magnetic field is applied to the target region, which heats the magnetic particles by magnetic hysteresis losses. For an ideal hyperthermia treatment, it is essential that the surrounding healthy tissue should not be damaged while the tumor cells are selectively destroyed. Thus, it was absolutely required for hyperthermia treatment planning to understand the thermal response occurring in biological tissues for a good quality of medical treatment [3].

The most efficient method to know the thermal response in biological tissue is experiment, but a complete experiment is difficultly performed for the variety of tissues and complexity of the physical and biochemical processes. Relatively, the analysis and modeling of the underlying thermal mechanisms becomes important. The classical Fourier's law, implying an infinitely fast propagation of thermal signal, is always employed to study the thermal behavior for the majority of practical applications. However, the literature [4,5] indicated that biological tissues need a relaxation time to accumulate enough energy to transfer to the nearest element in heat transfer. Thus, the related researchers proposed various non-Fourier bioheat transfer models to model the complex thermal behavior in human body [6–10]. The literature [11,12] further indicated that the whole anatomical structure of biological tissue can be split into vascular region and extravascular region and be treated as a fluid saturated porous medium. Convective heat transfer between the blood and tissue and blood perfusion makes that the blood temperature differs from the tissue temperature and varies. Heat transfer in living biological tissue should be non-equilibrium is much more realistic than equilibrium heat transfer assumptions [13]. As a result, the two-temperature models based on volume average to the local instantaneous governing equation for blood and tissue were proposed [11,12]. In advance, Roetzel and Xuan [13] splitted the blood flow into arterial and venous components. Based on one of the two-temperature models, Zhang [14] derived the generalized dual-phase-lag (DPL) model of bioheat transfer which the phase lag times are not independent and depend on the properties of blood and tissue, interphase convective heat transfer coefficient and blood perfusion rate.

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Nomenclature

a	length of tissue, m
c	specific heat of tissue, J/kg·K
c_b	specific heat of blood, J/kg·K
C_{sb}	coefficient, $C_{sb} = \rho c/(\rho_b c_b)$
f	parameter defined in Eq. (A3)
G	coupling factor between blood and tissue, W/m^3·K
H	new dependent variable, $H = r(T - T_{b0})$
$\bar{H}$	Laplace transform of H
k	thermal conductivity, W/m·K
K	parameter defined in equation (A4)
K_{sb}	coefficient, $K_{sb} = k/k_b$
ℓ	distance between two neighboring nodes, m
n	total number of nodes
P	power density, W/m^3
q_m	metabolic heat generation, W/m^3
q_r	spatial heating source, W/m^3
r	space coordinate, m
R	radius of tumor, m
s	Laplace transform parameter
t	time, sec
T	temperature of tissue, °C
T_B	new dependent variable, $T_B = T_b - T_{b0}$
T_b	arterial temperature, °C
T_0	initial temperature of tissue, °C
T_{b0}	initial temperature of blood, °C
w_b	perfusion rate of blood, m^3/s/m^3

Greek symbols

ε	porosity
λ	parameter defined in Eq. (A2)
ρ	density, kg/m^3
ψ	volume fraction of magnetic particles
τ_p	phase lag of spatial heating source, s
τ	relaxation time, s

Subscripts

b	blood
eff	effective
i	node number
j	number of sub-space domain
k	number of layer

The generalized dual-phase-lag model of bioheat transfer has been applied to simulate thermal response in the laser-irradiated tissues [15–19]. The present work would use the generalized dual-phase lag model of bioheat transfer to predict the behavior of temperature response in living tissue during magnetic tumor hyperthermia treatment. During hyperthermia therapy, the blood temperature was always regarded as a constant and the blood perfusion started up a negative feedback mechanism and transfers heat from heated domain to normal domain, and then changes the situation of temperature distribution [15–19]. In fact, the blood temperature undergoes a transient process before onset of equilibrium. For more fully exploring the heat transfer behavior in perfused tissue, this paper takes the transient effect of blood temperature into account based on the blood temperature model presented in the literature [14]. In other words, this work needs to analyze the bio-heat transfer problem with simultaneous equations in a two-layer concentric solid tissue.

There are mathematical difficulties, essentially, in solving heat transfer problem in solid spherical medium for geometry effect and singular point. The present problem raises the mathematical difficulties for the complexity of interfacial boundary conditions, the terms of relaxation times, and the transient blood temperature. Thus, a hybrid numerical scheme based on the Laplace transform, change of variables, and the modified discretization technique in conjunction with the hyperbolic shape functions [18,20] is extended to solve the present problem. In addition, porosity and coupling factor are particularly important to the generalized DPL model, the effect of the difference in porosity and coupling factor between tumor and normal tissue on the thermal response must be investigated. The papers [21,22] also show that the heating behavior in tumor hyperthermia is a function of tumor size. Therefore, the exploration for the influence of size of tumor on the temperature evaluation is involved. During hyperthermia therapy, a temperature within the tumor between 42 °C and 46 °C could ablate the tumor without damaging surrounding healthy tissues [21–23], so the suitable power dissipation is estimated with the inverse concept. In order to explore the effect of blood temperature, the results with the transient blood temperature are presented and compared with those with the constant blood temperature.

2. Mathematical formulation

In magnetic hyperthermia, magnetic particles are injected into and homogenously distributed in a small tumor surrounded by the normal tissue. The small tumor is regarded as a solid sphere with the radius R [1,24–25] and becomes a heat source of constant power density P in the small tumor for excitation of alternating magnetic field. For $t > 0$, heat symmetrically transfers in the radius direction. The temperature distribution in the tumor ($0 \leqslant r \leqslant R$) and normal ($R \leqslant r \leqslant \infty$) tissues is the function of the distance r from the center of the sphere and time t. The present work explores the thermal behavior in this system with the generalized dual-phase -lag bioheat transfer equation.

Zhang [14] derived the generalized dual-phase-lag bioheat transfer equation with tissue temperature as sole unknown from the two-temperature model [12]. In the spherical coordinate system, the 1-D form of the generalized dual-phase-lag bioheat transfer equation with constant thermal parameters can be written as

$$\left(1+\tau_q\frac{\partial}{\partial t}\right)(\rho c)_{eff}\frac{\partial T}{\partial t} = \left(1+\tau_T\frac{\partial}{\partial t}\right)k_{eff}\frac{1}{r^2}\frac{\partial}{\partial r}\left(r^2\frac{\partial T}{\partial r}\right) + G(T_b - T) + \left(1+\frac{\varepsilon\rho_b c_b}{G}\frac{\partial}{\partial t}\right)[(1-\varepsilon)q_m + q_r] \tag{1}$$

where

$$(\rho c)_{eff} = \varepsilon\rho_b c_b + (1-\varepsilon)\rho c \tag{2}$$

and

$$k_{eff} = \varepsilon k_b + (1-\varepsilon)k \tag{3}$$

are effective heat capacity and effective thermal conductivity, respectively. ρ, c, k, and T denote density, specific heat, thermal conductivity, and temperature in two regions. Subscript b is referred to blood. q_m is the metabolic heat generation and is only a function of r in the present problem. The spatial heating source q_r is defined as $q_r = Pu(t)$, where $u(t)$ is a step function and P is the power density. T_b is the arterial temperature. G is a lumped convection–perfusion parameter and is referred to as coupling factor between blood and the tissue. ε is a proportional rate. This equation takes the effects from the blood flow, thermal diffusion and the local thermal