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Analysis of heat transfer and burn damage in a laser irradiated living tissue with the generalized dual-phase-lag model

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ABSTRACT

For considering the non-equilibrium effect, the present paper uses the generalized dual-phase-lag model of bioheat transfer to describe the thermal behavior in a laser irradiated living tissue. The thermal damage is also estimated with the Arrhenius equation. The hybrid application of the Laplace transform and the modified discretization technique is employed to solve the present problem. The non-reduction of the generalized dual-phase-lag model to the Pennes equation is further explored. The effects of that the phase lag times depend on the porosity, heat capacities of blood and tissues, coupling factor, and the ratio of thermal conductivity of tissue and blood are taken into account. The results show that the generalized DPL bioheat transfer equation cannot reduce to the Pennes and classical DPL bioheat transfer equations, even with the effect of spatial heating source. Porosity and coupling factor are particularly important to the generalized DPL model. The relation between the phase lag times would demonstrate the characteristics of thermal response in tissue.

1. Introduction

For knowing the thermal response in biological tissue, experiment is the most essential method. However, it is difficult to perform a complete experiment due to the variety of tissues and complexity of the physical and biochemical processes. The analysis and modeling of the underlying thermal mechanisms have been relatively important to explore the medical treatment problems. The Pennes bio-heat transfer equation is commonly used to simulate thermal behavior in biological bodies for simplicity and validity. However, it was developed on the basis of the Fourier law which depicts an infinitely fast propagation of thermal signal. Actually, a relaxation time is needed to accumulate enough energy to transfer to the nearest element in heat transfer in biological tissues [1,2]. With reference to the facts, various non-Fourier bio-heat transfer models were proposed by researchers to model the complex thermal behavior in human body and have their corresponding strengths and limitations [3–7].

Nakayama and Kuwahara [8] and Xuan and Roetzel [9] claimed 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. The blood temperature differs from the tissue temperature and varies due to convective heat transfer between the blood and tissue and blood perfusion. In other words, heat transfer in living biological tissue should be non-equilibrium is much more realistic than equilibrium heat transfer assumptions [10]. Therefore, the two-temperature models based on volume average to the local instantaneous governing equation for blood and tissue were proposed [8,9]. Roetzel and Xuan [10] further proposed a more refined model that splits the blood flow into arterial and venous components. According to a two-temperature model, Zhang [11] developed the generalized dual-phase-lag bio-heat transfer equation. The phase lag times in the equation are not independent and depend on the properties of blood and tissue, interphase convective heat transfer coefficient and blood perfusion rate. The generalized dual-phase-lag bioheat transfer equation has been applied to simulate thermal response in a laser-irradiated tissue [12–14]. Narasimhan and Sadasivam [15] also used it to describe the thermal behavior in human eye during retinal laser irradiation.

Afrin et al. [12]. and Hooshmand et al. [14] have made the conclusion that the generalized DPL equation can reduce to the classical Pennes heat conduction equation when the phase lag times of temperature gradient (τ_T) and heat flux vector (τ_q) are both zero. Though the classical DPL equation of bio-heat transfer can be reduced to the Pennes bioheat transfer equation for $\tau_q = \tau_T$ [16], the

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Nomenclature

C	specific heat of tissue, J/kg K
c_b	specific heat of blood, J/kg K
C_0	protein concentration in the normal tissue
C_d	concentration of denaturated protein
f	parameter defined in Eq. (A3)
g	anisotropy factor
G	coupling factor between blood and tissue, W/m^3 K
H	new dependent variable, $H = T - T_{bi}$
$\widetilde{H}$	Laplace transform of H
k	thermal conductivity, W/m K
ℓ	distance between two neighboring nodes, m
n	total number of nodes
q_0	laser density, W/m^2
q_m	metabolic heat generation, W/m^3
q_r	spatial heating source, W/m^3
s	Laplace transform parameter
t	time, s
t_1	onset time of laser exposure, s
t_f	evaluated time of thermal damage, s
t_p	laser exposure time, s
T	temperature of tissue, °C
T_B	new dependent variable, $T_B = T - T_{bi}$
T_b	arterial temperature, °C
T_i	initial temperature of tissue, °C
T_{bi}	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)
ϕ_{in}	incident laser irradiance, W/m^2
μ_a	absorption coefficient, cm^{-1}
μ_s	scattering coefficient, cm^{-1}
μ_s'	reduced scattering coefficient, cm^{-1}
ρ	density, kg/m^3
τ_p	phase lag of spatial heating source, s
τ_q	phase lag of heat flux, s
τ_T	phase lag of temperature gradient, s
Ω	damage parameter

Subscripts

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

study [13] does not agree with the conclusion of Refs. [12,14]. However, the heating source dealt with in Ref. [13] is boundary heating, not space-dependent. At the same time, the calculated results in Ref. [12] show that the heat conduction described by the generalized DPL bioheat transfer model would differ from the classical Fourier's heat conduction. In addition, the literature [14] regarded the phase lag time of spatial heating source as the same as the phase lag time of heat flux. Actually, the phase lag time of spatial heating source differs from the phase lag time of heat flux in the generalized DPL equation derived by Zhang [11]. As stated by Welch [17], the development of bio-heat models is essential for further improving the thermal treatment methods, for pre-planning purposes, for on-line monitoring and decision support as well as for evaluation of the extent of thermal damage. Therefore, it is worthy to further explore the characteristics of the generalized DPL bioheat transfer model.

For the purpose of comparison, the present paper would do an extension study to the bioheat transfer problem in a laser-irradiated living tissue analyzed in the literature [12]. The scattering phenomenon of laser light propagation was regarded as being obvious and meaningful, so laser heating serves as a spatially-varied body heat source that heats up the tissue. The present paper uses the generalized dual-phase-lag bioheat transfer equation to estimate the tissue temperature for considering the effect of non-equilibrium heat transfer. And then, the thermal damage is predicted with the Arrhenius equation, which was accepted and used to evaluate the damage parameter [18].

The literature [12] regarded the phase lag times as independent parameters. However, Zhang [11] clearly stated that the phase lag times depend on the porosity, heat capacities of blood and tissues, coupling factor, and the ratio of thermal conductivity of tissue and blood. The present work takes this dependence effect into account for discussing whether the values of the phase lag times can be arbitrarily determined in calculation. The effects of the relationship between the phase lag times of spatial heating source and heat flux were ignored. This work would pay attention on them. The effect of spatial heating source and the results in Ref. [12] would also be inspected. The effects of the coupling factor between blood and tissue and porosity on the thermal response are investigated. Some mathematical difficulties must be overcome for solving the present problem. A suitable and efficient method is required for the accurate and stable solutions. A hybrid numerical scheme based on Laplace transform is employed for solving the present problem.

2. Problem formulation

Consider a broad laser beam with a uniform irradiance (ϕ_{in}) is applied normally to a finite slab of biological tissue with a thickness of L at time $t = 0^+$. When the spot size of the broad laser beam is much larger than the thickness of the thermally affected zone for the time period of interest, a 1-D model would be sufficient for analyzing the thermal response of the heated medium [12,19].

In order to consider the effect of micro-structural interactions in the process of heat transport, the classical dual-phase-lag model of bioheat transfer is proposed, based on equilibrium heat transfer assumption, as:

$$\left(1+\tau_T\frac{\partial}{\partial t}\right)k\nabla^2 T=\left(1+\tau_q\frac{\partial}{\partial t}\right)\left[\rho c\frac{\partial T}{\partial t}-w_b\rho_b c_b(T_b-T)-q_m-q_r\right] \tag{1}$$

where k, ρ, c, and T denote the conductivity, density, specific heat, and temperature of tissue. t is the time. c_b and w_b are, respectively, the specific heat and perfusion rate of blood. q_m is the metabolic heat generation and q_r is the heat source for spatial heating. T_b is the arterial temperature. τ_q means the phase lag of the heat flux and τ_T means the phase lag of the temperature gradient.

Zhang [11] derived the generalized dual-phase-lag bioheat transfer equation with tissue temperature as sole unknown from the two-temperature model. It can be written in 1-D form with constant thermal parameters as