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Thermal analysis of laser-irradiated tissue phantoms using dual phase lag model coupled with transient radiative transfer equation

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ABSTRACT

The present work is concerned with the development and application of dual phase lag (DPL) based heat conduction model for investigating the thermal response of laser-irradiated biological tissue phantoms. The developed heat transfer model has been coupled with the transient form of radiative transfer equation (RTE) that describes the phenomena of light propagation inside the tissue phantom. The RTE has been solved using the discrete ordinate method (DOM) to determine the 2-D distribution of light intensity within the tissue phantom, while finite volume method (FVM) based discretization scheme has been employed for solving the heat transfer model. The developed numerical model has first been verified against the results available in the literature. The results obtained in the form of temperature distribution through DPL model have been compared with conventional Fourier heat conduction model as well as with hyperbolic model. The effects of two phase lags terms in the form of relaxation times i.e. τ_T and τ_q associated with DPL model on the resultant thermal profiles have been investigated. Thereafter, the temperature distribution inside the biological tissue phantom embedded with optical inhomogeneities of varying contrast levels have been determined using the DPL-based model. Here the optical inhomogeneities represent the malignant (absorbing inhomogeneity) and benign (scattering inhomogeneity) cells present in an otherwise homogeneous medium. Results of the study reveal that the hyperbolic heat conduction model consistently predicts high temperature values and also the associated thermal profiles exhibit the largest amplitude of oscillations throughout the body of the tissue phantom. The DPL-based model results into relatively lesser oscillations due to the coupled effects of τ_T and τ_q. The conventional Fourier model, on the other hand, results into the lowest temperature values without any oscillations in the temperature profiles. The effect of the presence of varying nature of optical inhomogeneities is also brought out quite clearly using the developed DPL-based heat conduction model.

1. Introduction

Research on understanding the thermal response of laser irradiated biological samples has intensified in the past few years due to its importance in medical applications such as laser surgery [1,2], laser-induced hyperthermia [3–6], laser-based photo-thermal therapy [7] etc. Accurate prediction of thermal response of laser irradiated biological tissues is required for maximizing the efficiency of these techniques for selective destruction of abnormal cells (e.g. tumors; malignant and/or benign) with minimum thermal damage to the surrounding normal tissues. Of all the techniques available, laser-based photo-thermal therapy [7] has attracted considerable attention from a wide range of researchers and medical practitioners as it is considered to be the minimally invasive treatment method for the therapeutic applications. The technique is based on the concept of raising the temperature of the biological tissue sample up to a predefined threshold value that is high enough to destroy the cancerous cells present in an otherwise homogeneous tissue medium. The localized increase in the temperature of the disease-affected region is primarily due to the fact that the cancerous cells have significantly different optical properties (e.g. higher absorption coefficient) as compared to the rest of the homogeneous medium and hence absorb higher amount of light intensity compared to the surrounding cells.

The treatment efficiency of the laser-based photo-thermal therapy can be significantly improved by optimally selecting the laser parameters such as laser power to be deposited on the sample, wavelength range in which the laser needs to be operated as it would result in better control of selectivity of the target region etc. The advent of short pulse lasers have added a completely new dimension to the overall concept of photo-thermal therapy

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Nomenclature

c	speed of light in medium
c_v	specific heat of the tissue
G	incident intensity
I	intensity
I_b	blackbody intensity, $\sigma T^4/\pi$
k	thermal conductivity
M	number of discrete directions
q	heat flux
s	distance traveled by beam
$\hat{\mathbf{s}}$	unit direction vector
T	temperature
t	time

Greek symbols

$\Delta\Omega$	control angle
ε	emissivity
κ	absorption coefficient
σ	scattering coefficient or Stefan–Boltzmann constant
β	extinction coefficient, $\kappa + \sigma$
ω	weight in discrete direction m
Φ	scattering phase function
μ, ζ	direction cosines in x and y direction respectively
ρ	density of the medium
τ	relaxation time

Subscripts

b	blackbody or blood
w	wall/boundary
M	metabolic

due to their inherent advantages in terms of high laser power, repetition rates etc. Moreover, the time duration of laser-irradiation on the tissue volume can easily be made smaller compared to the thermal relaxation time. Hence, the subsequent rise in the tissue temperature as a result of the absorption of the incident laser power can be confined well within the localized tissue region before the thermal energy gets diffused to the surrounding normal cells. Furthermore, as part of the recent developments in this direction, plasmonic gold nanoparticles have found considerable attention in order to maximize the efficiency of these techniques for the thermal treatment of tumors [4–6]. These nanoparticles are characterized by strong resonance absorption and relatively weak scattering in the therapeutic window (≈0.6–1.4 μm) while the normal biological tissues exhibit high scattering and weak absorption characteristics within this wavelength range. These properties of plasmonic gold nanoparticles aid in localized treatment of cancerous cells with minimum possible thermal damage to the surrounding healthy tissues. The available literature shows that various shapes, sizes and materials of these nanoparticles have been used as the absorbing agent in the context of photo-thermal therapy [8,9].

For developing a comprehensive understanding of the thermal response of laser-irradiated biological tissues, the phenomena of laser light propagation through the body of the sample needs to be carefully addressed. Biological samples are generally composed of constituents that are absorbing (e.g. water content, hemoglobin, melanin etc.) as well as scattering (e.g. RBCs, cell membranes etc.) in nature and hence are classified as turbid medium. By virtue of this, the photons traveling through the laser-irradiated tissue medium undergo absorbing and multiple scattering events before emerging from its boundaries in the form of transmitted and/or reflected signals. The propagation of light intensity through a biological medium is generally mathematically modeled using the transient form of radiative transfer equation (RTE) [7,10–11] and a wide range of numerical techniques have been developed and applied by various researchers for solving this integro-differential equation in the past. These numerical models include discrete ordinate method (DOM), discrete transfer method (DTM), finite volume method (FVM), two-flux method etc. [12,13]. A comparative study of these methods presented by Mishra et al. [12] has demonstrated that the DOM is computationally most efficient of all the numerical methods available. It is pertinent to note here that in the context of photo-thermal therapy, while the RTE provides detailed insights into the time resolved and whole-field distribution of light intensity inside the body of the sample, it is equally important to understand the resultant temperature distribution for controlling the effectiveness of the technique. For example, a temperature increase of ≈5–10 °C has the potential of influencing the enzymes activities, bring changes in the blood flow and vessel permeability. On the other hand, temperatures of the order of ≈45–80 °C can lead to complete thermal denaturation of living biological cells [14].

The approach generally followed for the determination of temperature field distribution inside the laser-irradiated biological samples has been based on coupling the RTE with energy equation and a suitable heat conduction model [7,10–11,15]. The conventionally applied heat conduction models have primarily been based on the standard Fourier law of heat conduction [7,10]. However, it has been experimentally as well as numerically demonstrated by a group of researchers [11,16] that the temperature values as predicted by Fourier law of heat conduction do not accurately match with the experimental predictions. The discrepancy in the results has been attributed to the inherent assumption of infinite speed of thermal wave propagation through the body of the laser-irradiated biological sample, as made in the standard Fourier models [17,18]. Moreover, Fourier law shows limitations in applications wherein one employs short duration laser pulses, temperatures of the range of cryogenics, studying the thermal responses of non-homogeneous structures such as biological samples etc. [19–23]. In order to overcome these limitations of Fourier heat conduction, efforts have been made by a select group of researchers in developing non-Fourier heat conduction models that take into account the finite speed of light propagation through the biological samples which are inherently non-homogeneous in nature. In this context, Cattaneo [24] and Vernotte [25] modified the conventional Fourier law by introducing a phase lag term for the heat flux and the equation proposed is popularly termed as Cattaneo–Vernotte heat conduction equation or hyperbolic heat conduction equation expressed as

$$\vec{q} + \tau_q \frac{\partial \vec{q}}{\partial t} = -k\nabla \mathrm{T} \quad (1)$$

Though Eq. (1) can eliminate the paradox of infinite thermal propagation speed, it does not take into account the effects of micro-structural interactions in a non-homogeneous medium such as biological samples. In order to take care of these discrepancies, Tzou [26] modified the Cattaneo–Vernotte equation by introducing another term that introduces a phase lag in the resultant temperature gradient. This model is known as dual phase lag (DPL) model and can be expressed as