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Numerical investigation of the influence of pulsatile blood flow on temperature distribution within the body of laser-irradiated biological tissue phantoms

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ABSTRACT

The present work is concerned with the numerical investigation of the effects of pulsatile nature of blood flow in large size blood vessels on temperature distribution in laser-irradiated biological tissue phantoms. Phenomenon of light propagation through the tissue–blood vessel domain is modelled by transient radiative transport equation (RTE). Discrete ordinate method has been employed to solve transient form of RTE. In order to determine the temperature distribution inside the tissue–blood vessel domain, the solution of RTE has been coupled with the energy equation. Pulsatile nature of velocity profile has been employed at the inlet of the blood vessel. Navier–Stokes equations have been solved for determining the velocity field required to understand the effect of pulsatile blood flow on the temperature distribution. Relative influence of various parameters such as heart beat rate, time-averaged blood inlet velocity and size of the blood vessel on temperature distribution within the blood vessel has been presented. The numerical code developed as part of the present study has first been verified against the results available in the literature for the same operating parameters. Both homogenous as well as tissue phantoms embedded with optical inhomogeneities of different contrast levels have been considered. Results of the study clearly reveal the strong influence of pulsatile nature of blood flow on the resultant temperature distribution in the surrounding tissue region. The increase in temperature due to laser-irradiation is found to be relatively lesser in the presence of blood flow due to its convective cooling effects as compared to the rise in temperature achieved in the tissue phantom without blood vessels.

1. Introduction

Laser-based photo-thermal therapy has gained considerable attention in recent times for destroying the cancerous cells embedded in an otherwise homogenous biological sample. The technique is based on the principle of conversion of the energy of the incident photons into heat energy that causes cellular damage by thermal effects such as hyperthermia, evaporation and coagulation. Lasers have been widely adopted in the medical field by virtue of their properties such as high power, short duration capability, directionality, monochromatic nature etc. [1,2]. With the advent of short pulse lasers, it is possible to deposit the laser energy into a selected region in the tissue domain within a short duration of time. Hence, the rise in temperature due to the absorption of energy of photons can be restricted to the specified tissue volume before the heat gets diffused into the surroundings. In the context of therapeutic applications, it has been experimentally established that a temperature range of ≈50–60 °C is enough to selectively destroy the tumorous cells in the human body [3]. Hence it is important to achieve these values of threshold temperature limits over the affected regions and also to ensure the uniformity in the temperature distribution within the laser-irradiated region in the tissue. In practical applications, the living tissues have large blood vessels (e.g. aorta, arteries etc.) with diameter >500 μm [4] embedded within them. Blood flow within these large blood vessels convectively takes away the heat deposited during the process of laser-irradiation and hence the threshold temperature values required for complete tissue necrosis cannot be achieved due to the localized cooling effects for a given set of laser parameters.

The phenomena of light propagation through the laser-irradiated biological samples is mathematically modeled as the transient radiative transport equation (RTE), which is an integro-differential equation [5]. Over the years, various numerical models e.g. P_1 [6–8], P_N [8], Two-flux [8], discrete ordinate method (DOM) [8,9], finite volume method (FVM) [9], discrete transfer method

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Nomenclature

c	speed of light in medium
c_v	specific heat
D_h	hydraulic diameter
G	incident intensity
H_2–H_1	diameter of blood vessel
I	intensity
I_b	blackbody intensity, $\frac{\sigma T^4}{\pi}$
k	thermal conductivity
L	length
M	number of discrete directions
p	pressure
q	radiative heat flux
S	source function
s	distance traveled by a beam
$\hat{s}$	unit direction vector
u, v	velocity components
T	temperature
t	time
W	width

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 directionrespectively
ρ	density of the medium
μ	dynamic viscosity of blood

Subscripts

b	blackbody or blood
t	tissue
w	wall/boundary

Superscripts

$*$	dimensionless quantity
m	index for the discrete direction

(DTM) [9,10], etc. have been proposed and implemented by a range of researchers for determining the intensity distribution within the body of the laser-irradiated biological sample. While RTE provides useful information about the light propagation through biological samples and the resultant intensity distribution, it is important to understand the thermal behavior of the laser-irradiated sample and hence determine the temperature field distribution so as to develop a detailed understanding of the heat-affected zones in the surroundings of the targeted cells. Under conventional approach, the divergence of the radiative heat flux acts as the source term in the classical Pennes bio–heat transfer equation for determining the temperature distribution [11–15]. In this context, in one of the recent studies, Randrianalisoa et al. [16] employed the Monte Carlo-based ray-tracing (MCRT) method for quasi-steady radiative transfer as well as transient MCRT for unsteady radiative transfer problem for first determining the absorbed radiation power. The absorbed radiation power, thus determined, now acts as the source term in heat conduction equation for calculating the temperature variation during hyperthermia. It is to be noted here that authors in the reported study neglected the convective heat transfer with arterial blood and also neglected the metabolic heat generation in superficial tissues. With reference to the applicability of the classical Pennes bio–heat transfer equation in the context of photo-thermal therapy, it is to be noted that this equation is subjected to the assumption that the arterial blood temperature is almost uniform throughout the tissue while the venous blood temperature equals that of the local tissue temperature [6]. However, for large blood vessels with diameter in excess of ≈500 μm, the assumption of thermal equilibrium between the blood and the surrounding tissues is no longer valid [17–20] and hence the Pennes bio–heat transfer equation model cannot be applied for determining the resultant temperature distribution under these circumstances. In such large blood vessels, the energy transport equations for the blood and the surrounding tissues have to be considered separately.

In order to circumvent the above-mentioned limitations of Pennes bio–heat transfer equation, a group of researchers have attempted solving the energy equation separately in the blood vessel domain and the surrounding tissues. Of notable interests are the studies reported by Kolios et al. [21] wherein the authors investigated the cooling effects of large blood vessels embedded in uniformly heated tissues. Shih et al. [22] studied the cooling effects of thermally significant blood vessels on the extent of thermal lesion during thermal treatment. It was concluded that the short-duration and high-intensity heating scheme can completely reduce the cooling effect of the blood vessels that are 200 μm in diameter. However, in the case of large blood vessels, the authors reported that neither longer heating duration nor higher heating power density is sufficient for complete necrosis of the tumorous cells. The effect of single as well as two large countercurrent blood vessels on three-dimensional heat transport phenomena during laser-induced thermotherapy has been reported by Zhou and Liu [23]. In a recent study, Paul et al. [24] investigated the cooling effects of single and two large countercurrent blood vessels on temperature distribution in tissue-mimicking biological samples during laser-assisted photo-thermal therapy. The laser heat generation (modelled using Beer–Lambert law) was employed as the volumetric heat source into the energy equations for tissue, blood and the blood vessels. It is pertinent to note here that in both the above-mentioned studies by Zhou and Liu [23] and Paul et al. [24], the authors either employed the steady form of RTE or have modelled the light propagation through tissue-mimicking biological samples using the standard Beer–Lambert law. However, in view of the fact that photo-thermal therapy generally employs short-pulsed lasers (with pulse width of the order of picosecond to femtosecond) as the heating source, it is necessary to consider the transient form of RTE. Moreover, the biological samples are turbid in nature, hence the propagation of the light beam through a given biological medium cannot be completely defined by the standard Beer–Lambert Law which holds good for the collimated component of light alone and does not account for the diffuse component arising due to multiple scattering events.

It is also to be noted here that the authors in the above-mentioned studies have considered the parabolic velocity profiles for the blood at the entrance of the blood vessels. However, the cyclic nature of the heart pump creates pulsatile flow conditions in all the blood arteries [25–27], therefore it becomes imperative to study the effect of pulsation in the blood flow on the net energy exchange between the blood vessels and the surrounding tissues. In this context, some of the notable studies are those reported by Kim et al. [28] who studied the heat transfer in thermally developing region of a pulsatile flow (sinusoidal velocity profile) at the