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Analysis of photon transport in biological tissue and the subsequent heating effects

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ABSTRACT

Analysis of laser interaction with matter revealed the possibilities of many industrial and therapeutic applications. This research article discusses the theoretical aspects of laser beam interaction with biological tissues. It introduces the numerical analysis of photon distribution and transport in the tissue and its bio-thermal heating effects. The Monte Carlo method has been applied to simulate the variation of photon distribution and photon fluence with the radial distance from the point of interaction as well as laser powers and tissue thickness. For a specific wavelength, the variation of diffuse reflectance with the absorption coefficient was depicted for different values of the anisotropy factor. It has also been used to simulate the bio-heat transfer to obtain the temperature variation with the heating depth. On the other hand, finite difference method (FDM) has been applied to simulate the heating effect resulted from the incident laser beam on the tissue based on Penne's bio-heat equation combined with the obtained photon distribution and transport parameters from the MC method. The heating effect of the laser beam and hence the occurred thermal damage in the tissue was depicted. A linear relationship between the temperature and the rate of thermal damage has been manifested. This result can be used as a threshold reference for various medical applications of lasers.

1. Introduction

Lasers have been used as effective tools for both hyperthermia and thermotherapy applications where light energy is converted into heat upon absorption in the tissue. The investigation of photon interaction with matter can be studied through various optical and thermal processes that occur during the interaction. Some of those processes are utilized for tissue ablation (tumor cells), tissue cutting (surgical operations) and some others for spectroscopic purposes (i.e., inspection and/or imaging). In all cases, the optical modulation of the laser beam, its heating effect and its accompanied shockwave are prominent processes when employing laser light for certain applications. The effective energy coupling into the sample [1,2] represents the key issue for understanding the physics of laser–mater interaction and hence improving the efficiency and reliability of the interaction. Photon scattering and absorption are the most important processes during the laser irradiation on turbid media [3,4]. The amenable utilization of laser radiation for medical applications requires some considerations and safety precautions that might be incurred through exact modeling and simulations. However, the main challenge in the theoretical modeling and simulation is to facilitate the accurate application of energy for the desired medical application through a treatment planning [4–6]. Moreover, the laser beam parameters as well as the tissue properties need to be accurately optimized in the bio-heat model to get an optimum and precise design for high temperature therapy applications.

MC method is usually utilized for simulating random process and has been applied to light–tissue interactions under a wide variety of situations [7–10]. Investigation of Photon interaction with matter based on scattering and absorption is stochastic in nature and can be described using the MC method by appropriate weight absorption and scattering events [11]. Laser irradiation of biological tissue of homogeneous and layered geometries [11–18] has been effectively simulated using the MC method.

The application of the MC method to simulate the light transport in tissue is based on the radiative transfer energy. It involves calculations of photon propagation in turbid media where the

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photons are treated as particles. The phase and polarization of the light are usually not considered, which is justifiable due to the multiple scatterings of the photons [17–20]. In the simplest MC simulation, the photons are individually incident on the medium and their paths are traced until they get either absorbed or permanently scattered out of the region of interest [20–23]. The rules of photon propagation are expressed as probability distributions, which are based on the geometry and optical properties of the medium. The input parameters are the absorption and scattering coefficients as well as the scattering phase function of the tissue. The main advantage of MC simulations over the diffusion approximation is that MC does not require the condition $(\mu_s^{'}(\lambda) >> \mu_a(\lambda))$ and accurate results can be obtained close to sources and boundaries [21]. In addition to that, it is feasible in the modeling of complicated geometries and multi-layered tissues that have optical properties of spatial variation [22,23]. On the other hand, Finite difference method (FDM) is considered as one of the efficient numerical techniques for solving the bio-heat diffusion equation. FDM relies on the quantization of time and position space, based on physical parameters of the problem.

In this study, the numerical analyses have been considered to acquire information about laser interaction with the sample through; photon distribution and transport in the tissue, temperature distribution during and after laser irradiation as well as the variation of the rate of thermal damage with temperature. The dynamic processes of thermal responses to laser irradiation and variation of various interaction parameters have been optimized in accordance with some related experimental results reported in literatures [7,9,10,14]. A combination of Monte Carlo (MC) method and bio-heat finite difference method (FDM) has been applied to simulate the probabilistic photon distribution in the tissue and predict the heating effects (profile and threshold of damage).

2. Theory of light propagation in the tissue

In tissue optics, the scattering and absorption properties are strongly wavelength-dependent. They are of great research importance for the detection and localization of cancer tumors in a harmless manner. Fig. 1 shows the general configuration of the dynamic of light–tissue interaction and its important parameters. When the laser beam is incident on the tissue surface and due to the refractive index mismatch, it will experience some reflection according to the Fresnel reflection formula [2].

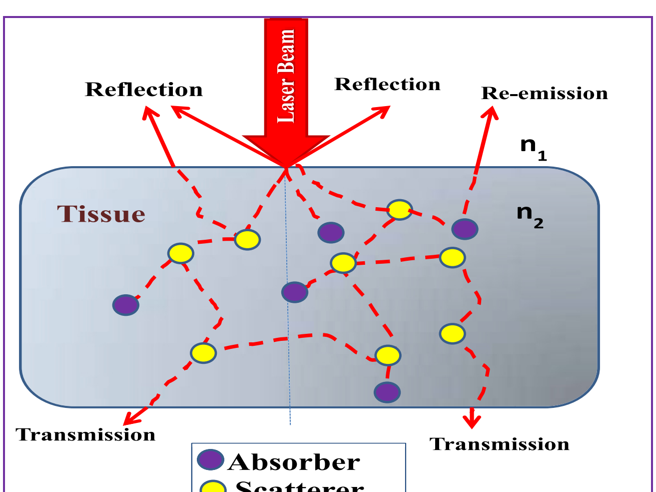

Fig. 1. Schematic representation for laser tissue-interaction.

The reflectivity should be taken into account when considering the actual laser beam intensity that propagates inside the tissue. The propagated beam will be attenuated due to the effect of many processes that may occur, but the dominant ones are the absorption and scattering processes.

2.1. Propagation of laser light in the tissue

The distribution of laser light in the tissue is dependent on both the optical properties of the tissue and the laser parameters. The propagation of absorbed and scattered light is described by the radiative transport equation (RTE) [3].

$$(s.\nabla)L(\vec{r},\hat{s},t) = -(\mu_a+\mu_s)L(\vec{r},\hat{s},t) + \mu_s \int_{4\pi} P(\hat{s},\hat{s}')L(\vec{r},\hat{s}',t)d\Omega' + Q(\vec{r},\hat{s},t). \quad (1)$$

The RTE is an integro-differential equation relates the gradient of radiance $L(\vec{r},t,\hat{s})[Wm^{-2}sr^{-1}]$ at a position r in a direction $\hat{s}$ to the losses owing to the absorption and scattering and to the gain owing to the light scattered from all other directions $\hat{s}'$ into the direction $\hat{s}$. The left-hand side of the RTE describes the rate of radiance change at a point indicated by $\vec{r}(x,y,z)$ in the direction $\hat{s}$. The gradient of radiance in the $\hat{s}$ direction can be equivalently expressed as a sum of temporal and spatial terms as

$$(S.\nabla)L(\vec{r},\hat{s},t) = \frac{1}{v}\frac{\partial L(\vec{r},\hat{s},t)}{\partial t} + \frac{\partial L(\vec{r},\hat{s},t)}{\partial s}, \quad (2)$$

where v is the light velocity in the medium. The first term on the right-hand side is the energy attenuated due to absorption and scattering. The total attenuation coefficient $\mu_t(\lambda)$ is given by

$$\mu_t(\lambda) = \mu_a(\lambda) + \mu_s(\lambda), \quad (3)$$

where $\mu_s(\lambda)$ is the scattering coefficient that represents the inverse of the mean path length between scattering events (m^{-1}). $\mu_a(\lambda)$ is the absorption coefficient that represents the inverse of the path length between photon absorption events (m^{-1}). It is given in terms of the extinction coefficient k' and light wavelength λ_0 as $[\mu_a(\lambda) = 4\pi nk'/\lambda_0\,]$ where n is the refractive index of the medium.

The second term is the energy increase due to the radiance from all other directions scattered into the direction $\hat{s}'$ about a solid angle $d\Omega$. It is known as the scattering phase function, which is the probability density function of the light scattering in the direction of the unit vector $\hat{s}'$ from an incident direction of the unit vectors $\hat{s}$. The last term is the source term, which is the radiance of the source at the location r in the tissue and in the $\hat{s}$ direction (W m^{-3} sr^{-1}). The other important parameters that need to be considered in characterizing the laser tissue interaction include: the albedo parameter (b), the optical penetration depth (δ) and the anisotropy factor (g). They are expressed as

$$b = \mu_a/(\mu_a+\mu'_s); \quad \delta = \left(3\mu_a\left[\mu_s'+\mu_a\right]\right)^{-1/2}; \quad g = \frac{\int_{4\pi} P(\hat{s},\hat{s}')(\hat{s}.\hat{s}')d\Omega'}{\int_{4\pi} P(\hat{s},\hat{s}')d\Omega'} \approx \int_{4\pi} P(\theta)\cos\theta d\Omega', \quad (4)$$

where $\mu'_s = (1-g)\mu_s$ is the reduced scattering coefficient and g is the averaged value of the cosine of the scattering angle θ (expected cosine of angles of scattering). It describes the variation of light in the direction of propagation due to scattering events. $P(\theta)cos\theta$, is a