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Estimating the best laser parameters for skin cancer treatment using finite element model

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ABSTRACT

Skin cancer is an intimidating disease which necessitates the presence of a non-invasive treatment. Laser-induced thermotherapy is one of the recent noninvasive modalities of superficial lesion treatment. Although of its promising effect, this method still needs more effort to be quantized. Many studies are being conducted for this purpose. Modeling and simulating the process of skin lesion treatment by laser can lead to the best quantization of the treatment protocol. In this paper, we provide finite element models for the treatment of skin cancer using laser thermal effect. A comparison between the effects of using different laser parameters of diode laser (800nm) and Nd:Yag laser (1064 nm) revealed that Nd:Yag laser can be used effectively foe skin cancer treatment specially with high intensities of about 10^6 w/m^2.

Keywords: FEM, laser-skin interaction, thermotherapy .

INTRODUCTION

One of the most intimidating diseases is skin cancer. Like other tumors, skin tumor should be investigated and treated as early as possible. Moreover, skin cancer necessitates a noninvasive treatment modality. This is not only because of the side effects of invasive methods, but also because of beauty purposes. A recent and promising treatment for many pathologies is by absorbed laser energy. When a laser beam is incident on a biological tissue, it may be reflected, scattered, absorbed and/or transmitted. The process of laser absorption is the key to effective laser use. Without absorption, there is little or no tissue response (except in the case of plasma-generated shock wave). The absorbed energy is transferred to some other molecules/group of atoms in the tissue that is to be altered or changed. Once this photon energy is absorbed, the energy has to produce some effect. Normally, we think of heat production as the result of laser absorption. However, heat is just one way that the energy of the photon can be dissipated. The increased temperature alters the tissue and brings the different biological effects that are used clinically[1]. [B. M. Achuer, Vectoria M. Vander Kam and Michael W. Berns " Lasers in Plastic Surgery and Dermatology",Thieme Medical Publishers, Inc. ,pp. 88-91, 992],[2]. Because of the smaller penetration depth and the very localized thermal effect of the laser-tissue interaction, laser-induced thermotherapy is very suitable for the early treatment of superficial cancer,[3].[4].

Now a days, Laser-Induced Thermo Therapy (LITT) is being used widely. Although of its promising results in many medical applications, the treatment procedure still needs more quantification.

Experimental as well as modeling studies are being conducted to determine a standard treatment protocols specially for cancer. The goal is always to kill the tumor cells with minimal damage to the normal tissue.

The authers in [5], provide method to model the thermal interaction in the skin by studying skin optics, light propagation, montecarlo simulation of laser beam and estimation of thermal effect. They used pulsed Nd:YAG laser. Pulsed duration range from 100 ms to 2 ms. They covered a very wide range from determination of laser beam and skin optical properties, treatment of light propagation in the tissue, thermal diffusion of heat to thermal effects prediction and thermal relaxation time. Their results revealed that using model-based parameter estimation, it appears that the relaxation time of the fast part (20 ms) is mainly determined by epidermis thickness, the relaxation time of the slow part (10 sec) by penetration depth and the ratio of amplitudes by the ratio of epidermis/dermis absorption coefficients.

(Review)

In this paper, we provide finite element models to investigate the laser parameters that can lead to effective skin cancer treatment while minimizing normal tissue damage. We consider different skin layers (epidermis, dermis and subcutaneous layers) and their properties instead of treating it as one layer. We also considered the case of superficial as well as subcutaneous tumors to build the geometrical models. Mathematical models were built based on the bioheat equation by which we could obtain an estimate for the temperature distribution inside the irradiated tissue. According to the temperature reached, the time of irradiation and the percentage of damage, we evaluate the evidence of the treatment.

A description of the models and the methods of analysis are provided in the following section. The results obtained from these models are presented and discussed in section III. We conclude our work and suggest future work in the last section of this paper.

MATERIALS AND METHODS

Mathematical modeling of the thermal effect of laser-tissue interaction can be done by studying the process of heat generation and how it can be transported through the tissue and raise its temperature. So, an estimation of the temperature of a point of interest (or generally, the heat distribution) is of great importance since it can lead to the prediction of certain thermal effect.

To get an estimate for temperature T at a point in a thermal system , we have to know what heat sources that might affect this temperature. For a thermal system, any alteration in the heat content of that system leads to an alteration of its temperature. This is given by the basic law of thermodynamics:

$$dQ = mcdT$$

$$\dot{Q} = mc\dot{T}$$

Where Q is the heat content of the system, m is its mass in kg, c is its specific heat in kJ/kgK and T is its temperature. If we consider the heat content in unit volume as q then:

$$\frac{\dot{Q}}{V} = \frac{mc}{V}\dot{T}$$

$$\dot{q} = \rho c \dot{T}$$

$$\dot{T} = \frac{1}{\rho c}\dot{q} \qquad (1)$$

The alteration in q may be due to gain of heat by external or internal sources and loss of heat by conduction, convection and radiation.

In case of biological tissue exposed to laser, the external source of heat q_l is due to light excitation, the internal source is due to the heat generated by metabolic activities per unit volume q_m, heat radiation could be negligible, convection heat q_{bp} is heat transfer due to blood perfusion per unit volume of tissue and conduction heat q_{cond} which is the considerable source of heat loss, is the primary mechanism by which heat is transferred to unexposed tissue structure [Niemz]. So, the temporal change of total heat content $\dot{q}$ is given by:

$$\dot{q} = \dot{q}_{cond} + \dot{q}_{bp} + \dot{q}_{met} + \dot{q}_l \qquad (2)$$

According to the equation of continuity, the temporal change of heat content due to conduction is determined by the divergence of heat flow j_Q:

$$\nabla . j_Q = -\dot{q}_{cond}$$

But the heat flow j_Q is proportional to the temperature gradient according to the general diffusion equation:

$$j_Q = -k\nabla T$$

Where k is the heat conductivity in $Wm^{-1}K^{-1}$

$$\nabla . j_Q = -\dot{q}_{cond} = -\nabla . k\nabla T$$

$$\dot{q}_{cond} = k\nabla^2 T$$

Which is the general homogenous heat conduction equation.

In the biological tissue, heat convection is done by the passage of blood through the vessels. The perfusion of blood through the tissue causing heat to change. The temporal change of heat content per unit volume of tissue caused by blood perfusion $\dot{q}_{bp}$ is first introduced by Pennes in 1984 and is given by:

$$\dot{q}_{bp} = \rho_b c_b \omega_b (T - T_a)$$

Where T_a is the arterial blood temperature and ρ_b and c_b are the density and specific heat of blood, ω_b is the blood perfusion (the volume of blood per unit volume of tissue per unit time)[PHAS 4886].

The external heating source in our case is by laser irradiation. Usually, the intensity of the laser beam used in medical applications is of Gaussian distribution and has an exponential decay along the direction of propagation z inside the tissue. The intensity profile is Gaussian during the laser pulse width and for simplicity, could be considered uniform with short pulses. Considering a cylindrical tissue construct, the optical intensity inside the tissue is given by:

$$I = I_o \exp(-\frac{2r^2}{w^2} - \alpha z) \exp(-8\frac{t^2}{\tau^2})$$

Where Io is beam intensity of the surface of the tissue, w is the beam waist, α is the absorption coefficient of the tissue and τ is the duration of laser application. With the propagation of photons inside the tissue, the heat deposited per unit area and time and in a thickness Δz is related to the light intensity gradient across this thickness by:

$$S(r,z,t) = \frac{I(r,z,t) - I(r,z+\Delta z,t)}{\Delta z}$$

in units of W / Cm^3.

As Δz approaches zero,

$$S(r,z,t) = \frac{\partial I(r,z,t)}{\partial z}$$

$$S(r,z,t) = \alpha I(r,z,t)$$

Which represents the temporal change of heat density due to light excitation $\dot{q}_l$ at certain point.

$$\dot{q}_l = \alpha I$$

The metabolic heat is a constant that can be given according to the activity and the situation of tissue being irradiated.

We can now write equation (2) as:

$$\dot{q} = k\nabla^2 T + \rho_b c_b \omega_b (T_a - T) + \alpha I + \dot{q}_{met}$$

From (1):

$$\rho c \dot{T} = k\nabla^2 T + \rho_b c_b \omega_b (T_a - T) + \alpha I + \dot{q}_{met}$$

$$\rho c \frac{\partial T}{\partial t} = k\nabla^2 T + \rho_b c_b \omega_b (T_a - T) + \alpha I + \dot{q}_{met} \qquad (3)$$

Equation (3) is known as the general bioheat equation which models the temperature changes trough the optically-irradiated biological tissue. Solution of this equation could be done numerically. Also, the finite element modeling software packages like COMSOL provide accurate solutions for such problems.

During the design of the geometrical models, we took into consideration the different layers of skin (epidermis and dermis), the subcutaneous layers and the underneath tissue, with the tumor being inside the subcutaneous or inside the underneath tissue. Because different layers have different optical properties, we consider them with their relative thickness to get accurate results.. We considered a cylindrical construct of the tissue with the boundary conditions at which the temperature gradient ∇T and hence the heat flow j_Q is zero. Assuming the Gaussian distribution of optical energy along the radius of the cylinder, we took a section of tissue along the optical axis to represent the geometrical model as shown in figure1.

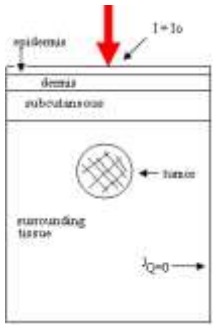

Figure 1: the geometrical model of skin cancer-laser treatment.

In this study, we used a laser beam of exponential intensity distribution with the tissue surface and exponential decay along the direction of propagation (optical axis) with an initial intensity I = Io on the tissue surface. Since neither macromolecules nor water strongly absorb in the near IR, a "therapeutic window" is delineated between roughly 600nm and 1200 nm. In this spectral range, radiation penetrates biological tissues at a lower loss, thus enabling treatment of deeper tissue structures Niemz in [3]. The Nd:Yag (1064nm) laser is used in this study because it is known for its effectiveness in skin cancer treatment. We compared the usage of diode laser (800 nm) with the Nd:Yag as diode laser is also in the therapeutic window and known for its low cost. We applied the models with different durations of exposure and different initial intensities.

The geometrical and the mathematical models together with the parameters needed to solve the bioheat equation were fed to COMSOL multiphysics program to get the temperature distribution and investigate the thermal effect. Table 1 and 2 show the properties of different tissues and blood used in this work.

TABLE 1: PARAMETERS REQUIRED BY THE BIOHEAT EQUATION FOR DIFFERENT TYPES OF TISSUES USED IN THIS WORK.

Parameter	Epidermis	Dermis	Subcut.	Tissue
Thermal onductivity k in w/(m.K)	0.23	0.45	0.19	0.5
Density ρ in kg/m³	1200	1200	1000	1050
Specificheat c in J/(kg.K)	3863	3573	2675	3600
Blood perfusion rate ωb in (1/s)	0	1.25x10-3	1.25x10-3	6x10-3

TABLE 2: BLOOD PARAMETERS REQUIRED BY THE BIOHEAT AND USED IN THIS WORK.

Parameter	Value
Arterial blood temperature in Ta in (K)	310.15
Density of blood ρb in (kg/m³)	1000
Specific heat cb of blood in (J/kg.k)	4200

RESULTS AND DISCUSSION

The results of processing a number of eighteen models were obtained and investigated. In these models we used diode and Nd:Yag lasers with three different values of intensities (10^4, 10^5 and 10^6 w/m^2), each wavelength and intensity were used with three different exposure times (300, 100 and 6 sec).

Form the obtained results we found that diode laser is not feasible at all with intensities lower than 10^6w/m^2. Since the max temperature the tumor reached was 43º within a period of 300 second. This is very far from the temperature of irreversible cell damage which must be more than 52º for 300 second exposure time. With the higher intensities of 10^6 w/m^2, it is also not recommended to use diode laser. Since the tumor temperature reached 69º but with irreversible damage of surrounding normal tissue.

On the other hand, the Nd:Yag laser was shown to be highly effective in skin cancer treatment. It caused the irreversible tumor cell damage in a period of 6 second only while maintaining the surrounding tissue in the reversible damage regions.

Thermal distribution graphs of the models with (Diode, 300 sec, 10^6 W/m^2) and (Nd:Yag, 6 sec, 10^6 W/m^2) are shown in figure 2 and b. Figure 3 shows the temperature against the time for the same two models. It is clear from the figure 2 and 3 that diode laser causes irreversible normal tissue damage because of the several minutes of hyperthermia.

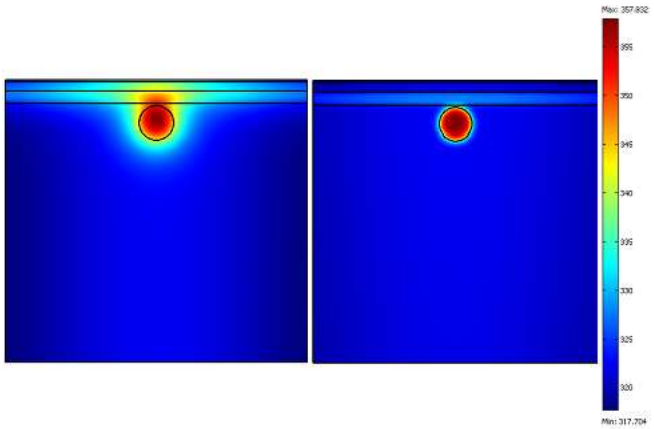

Figure 2: Thermal distribution graphs of the models

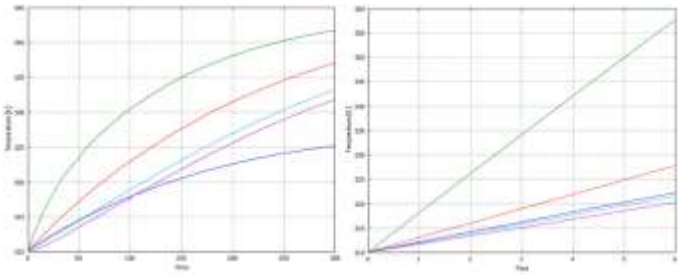

Figure 3: temperature against the time for the same two models

CONCLUSION

At the end, we conclude that not all lasers of wavelengths lying within the therapeutic window can be used for thermal treatment of tumor. This is because some wavelengths amy require a long exposure time to raise the tissue temperature to the desired value. This in turn must cause an irreversible damage of tissue. Pulsed Nd:Yag laser was shown to be superior in thermal treatment of tumor as it the tumor temperature could be raised to the required temperature in a short time which cause irreversible damage of tumor wile maintaining the normal tissue.

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