Peculiarities in the Distribution of Temperature under the Influence of a Laser Beam in a Multilayered Medium

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Abstract: *The term thermal interaction refers to a large group of interaction types where the local temperature rise is an important parameter. Temperature is the main parameter for all thermal interactions between laser beam and tissues. In addition, in order to predict the thermal reaction in tissues, a model of distribution of temperature in a thermal reaction is necessary. Very often in biological tissues, several thermal effects are observed depending on the properties of the laser.*

Keywords: *Biomedical technologies, Lasers, Mathematical model.*

Introduction

During the process of interaction of the laser beam with bio-tissues, heating is observed. Heat generation depends on parameters such as beam energy, treatment duration and absorption coefficient of tissues. The heat transfer is characterized only by the thermal properties of the tissues, which are thermal conductivity and thermal intensity. Conclusively the thermal effect depends on the type of the treated tissues and on the temperature at which they are able to be heated [5, 11].

The term thermal interaction refers to large groups of interaction types where the local temperature rise is an important parameter. Depending on the duration of the treatment and the maximum temperature, values achieved in tissues different effects can be identified such as: hyperthermia, coagulation, evaporation, carbonization (charring) and melting can be observed [7, 19].

Study and modelling of lasers in thermal effects

Three different effects contribute to the treatment result:

- ✓ Direct destruction of cells;
- ✓ Mediated destruction of cells due to the destruction of capillaries;
- ✓ Activation of heat and other biological processes.

In this manner, when heating in the range of 42-44 °C, tumours show signs of slight increase in sensitivity in comparison to normal tissue [8, 14].

For instance, the temperature at which tissue coagulation starts is not a constant value but rather depends on the duration of the treatment. This dependency on the constants of the reactions on the temperature can be demonstrated by means of the Arrhenius equation [2, 9]:

$$-\ln\frac{C(t)}{C_0}=\Omega(t)=\int_0^t\{A\exp(-E_a/RT(\tau))\}d\tau, \quad (1)$$

where C_0 is the initial concentration of molecules or cells,

$C(t)$ – the concentration at a given time t,

A – speed constant of the chemical reaction of protein denaturation,

E_a – activating energy,

R – universal gas constant,

$T(\tau)$ – the absolute temperature at any moment of time.

The degree of thermal damage can be determined by the expression:

$$\frac{C_0-C(t)}{C_0}=1-\exp(-\Omega),$$

where the heating duration is in conformity with the fixed temperature T_{cr} and the denaturation value (the relative concentration of damaged molecules) goes up to the value of $1-e^{-1}\approx 0{,}63$ [1, 10]. This temperature is called denaturation temperature.

When a laser beam treated with a certain wavelength effects tissues of high water content the effect of thermal destruction takes place. The kinetics of photo and thermal decomposition of biopolymers under laser treatment restricts the relatively fast diffusion processes as migration of free gases and carbon coagulation [4, 18].

The time characteristics needed for carbonization can be assessed through

$$t_c=l^2/D,$$

where l is a characteristic of the diffusion path,

D – diffusion coefficient that increases rapidly with the increase of the temperature:

$D(T)=D_0\exp\left(-U/kT\right)$.

Here U is the activation energy,

k – Boltzmann's constant,

D_0 – coefficient.

For the process of carbonization to occur the duration of the laser treatment must be more than t_c, i.e., the condition $\tau\geq l^2/D$ must be achieved. Particularly, the duration of the treatment must not be too short and temperature increased.

The thermal action of the laser beam can be used only in cases when the power density is ≥ 10 [W/m^2] for continuous or pulse radiation with the impulse duration exceeding 1 μs. Carbonization and drying can be practically achieved with a treatment from any kind of laser if sufficient power density and exposure duration are provided [9, 14].

Temperature is the main physical quantity that characterizes all thermal treatments of light with tissues. Quite often not only one but several thermal effects are induced in bio-tissues, depending on the parameters of the laser. They can have different effects ranging from the carbonization at the tissue surface or hyperthermia a few millimetres inside the tissue. But in all cases the objective is to achieve only one specific effect. Therefore, a careful evaluation of the parameters of the laser is substantial [10].

In order for the thermal reaction process to be predicted, a model of temperature distribution inside the tissues should be constructed. The spatial magnitude and the degree of damage of tissues depends on the parameters of the laser and on the optical tissue properties. When describing the spread of heat in tissues, thermal power, heat conductivity and tissue density are very important. The possible thermal processes are illustrated in Fig. 1. The location and spatial extent of each thermal effect depends on the temperature of the biological tissue during and after the laser treatment.

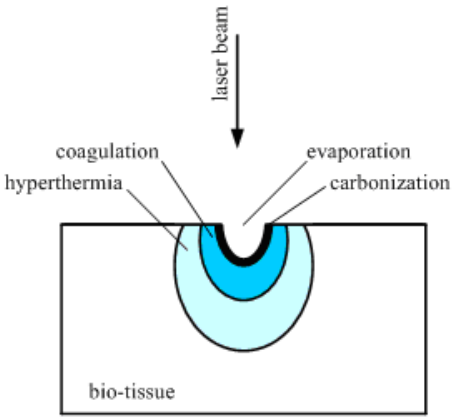

Fig. 1 Location of thermal effects in biological tissue

A. Heat generation

Generation of heat inside biological tissues during laser treatment is a result of the quantity of radiation that is absorbed in the tissue. In a non-scattering medium, the local heat deposition per unit area for a unit of time is assessed and can be calculated with the help of the formula:

$$S(r,z,t) = \mu_a I(r,z,t), \tag{2}$$

where z is the optical axis,

r – distance from the optical axis,

t – time,

μ_a – absorption coefficient,

$I(r,z,t)$ – local intensity.

Then, the function of the heat source $S(r,z,t)$ in the exposed tissues is a function of the coefficient of absorption and the local intensity. Since μ_a depends highly on the wavelength, this dependency refers to $S(r,z,t)$. If no phase transitions (evaporation, melting) or alterations of tissues (coagulation, carbonization) occurs, the change in the heat quantity dQ causes a linear change of temperature dT in accordance with the basic laws of thermodynamics [12]:

$$dQ = mcdT, \tag{3}$$

where m is the mass of the tissues,

c – specific heat capacity of the medium, [J/kgK].

B. Heat transfer

Within a closed physical system the relation between temperature and heat content is shown through a mathematical Eq. (3). In a real laser treatment the laser beam with the substance, the heat losses based on the heat conduction, heat convection and heat radiation should be taken into consideration. In most cases of laser treatment, the last two effects can be ignored. A typical example of heat convection transfer in tissues is the heat transfer through blood flow.

Temperature is the main physical quantity that characterizes the complete thermal interaction of light with tissues. The main task of the analytical theory of heat conductivity is the identification and scrutiny of the spatial-temporal alteration in temperature:

$$T = f(x,y,z,t),$$

where x,y,z is the spatial rectangular coordinate system, t – time.

The combination of temperature values of all points in an area at a certain moment of time t is called a temperature field. Moreover, it is a scalar quantity. If temperature is a function only of coordinates, the field is stationary; if temperature also depends on the time, then the field is not stationary.

The differential equation of thermal conductivity connects the spatial distribution of temperature to its change in time and acquires the following equation [15]:

$$\rho \cdot c \frac{\partial T(\vec{r},t)}{\partial t} = \nabla(k \nabla T(\vec{r},t)) + S(\vec{r}) \tag{4}$$

$$S(\vec{r}) = \mu_a \cdot \varphi(\vec{r}) \cdot \frac{E_0}{\tau_p},$$

where ρ is tissue density in [kg/m^3],

c – specific thermal capacity of the medium in [J/kg·K],

t – time,

$k = a_t \cdot \rho \cdot c$ – thermal conductivity in [W/m·K],

a_t – temperature conductivity in [m^2/s],

$S(\vec{r})$ – volume density of the power of the heat source in the medium [W/m^3],
μ_a – absorption coefficient,
$\varphi(\vec{r})$ – full illumination in point $\vec{r} = (x, y, z)$,
E_0 – the radiant energy density in [J/m^2].

This equation shows the change of the substance energy in a simple volume. It depends on the heat accumulated due to thermal conductivity and the heat released in a basic volume due to internal sources [1, 17].

If the model contains a couple of layers with different thermal-physical properties, then the Eq. (4) should be solved for each layer individually. In these areas with additional (that are not affected by the laser exposure) sources or heat consumers (absorbers) the heat calculations can be achieved through Eq. (4). In the areas where such sources are present or heat absorbers additional elements should be added to the heat conductivity equation.

Situations (for example, human skin) where blood is considered as an additional source are observed. Regarding areas where blood vessels do not exist anatomically, Eq. (4) is solved. In areas where blood vessels are present heat sources are added due to blood flow (if the calculations are made providing that blood is evenly distributed in the whole volume layers). Then the equation for heat conductivity acquires the following aspect [6, 13]:

$$\rho \cdot c \frac{\partial T(\mathbf{r},t)}{\partial t} = \nabla(k \nabla T(\mathbf{r},t)) + Q(\mathbf{r}) + S_{blood}(\mathbf{r},t,T). \quad (5)$$

Blood can absorb as well as release heat depending on how its temperature corresponds to the temperature of the surrounding tissue. The respective addend $S_{blood}(\mathbf{r},t,T)$ is the whole volume of blood flow or source of energy and can be written in the following way:

$$S_{blood}(\mathbf{r},t,T) = \rho \cdot c[\rho_b \cdot f(t,T) \cdot (T_{blood} - T(\mathbf{r},t))], \quad (6)$$

where ρ_b is blood density,
T_{blood} – blood temperature,
$f(t,T)$ – is the density of the blood flow in the tissues and it is measured in [ml/100 g·min] or [cm^3/s·g] [4], it is the volume of blood that is transferred for a second per gram of the respective tissue.

In stationary conditions, the density of blood flow is equal to f_0 and it is different in derma and in fats. Despite that, the stationary density of blood flow depends on the temperature, and $f_0(T)$.

In non-stationary conditions and under the exposure of light on skin, $f(t,T)$ is not equal to $f_0(T)$, where T is the momentous temperature of the tissues. As it is known at the change of temperature, the temperature of the blood flow does not change immediately but with a delay of t_{delay} = 60-90 s. The dependency of temperature on blood flow density is taken into account by means of the equation [15]:

$$\frac{\partial}{\partial t} f(t,T) = \frac{f_0(T) - f(t,T)}{t_{delay}}. \quad (7)$$

If we consider that temperature is a constant and does not change in time, then we can assume that the blood going through the area of the laser treatment will also slightly heat up.

The Eq. (5) shows the phenomenon of heat conductivity in general. The highlight of the process in question and its description requires the identification of the final conditions that should cover all geometrical, physical, initial and boundary conditions. The first two types of conditions comprise the geometry of the problem and the properties of the object. It is also important to consider the initial and boundary conditions [16].

The initial conditions are needed when reviewing non-stationary processes and influence the distribution of temperature inside the object at the initial moment of time. In general, the initial condition for the analytical equation of heat can be written in the following way (provided that $t = 0$): $T = T(x, y, z)$. At an even distribution of temperature in the body, the initial condition is simplified (if $t = 0$): $T = T_0 = const$.

According to Newton-Richman law, the amount of heat released on a unit of surface in a unit of time is proportional to the temperature difference between the surface of the body T and the temperature of the environment T_0, $(T > T_0)$:

$$q = \alpha(T - T_0), \quad (8)$$

where α is the coefficient of proportionality, which is also called coefficient of heat transfer [W/m^{2}K] and characterizes the intensity of heat exchange between the surface of the body and the medium. This coefficient in numbers is equal to the heat quantity released or absorbed from a unit of surface in a unit of time when the temperature difference between the surface of the body and the medium is equal to one degree.

Pursuant to the law of conservation of energy, a quantity of heat dispersed per unit of surface and per unit of time as a result of heat release should be equal to the quantity of heat supplied per unit of surface and per unit of time as a result of heat conductivity of the internal volume of the body, hence:

$$\alpha(T - T_0) = -k\left(\frac{\partial T}{\partial n}\right)_S, \quad (9)$$

where n is the normal to the surface of the body S.

The condition of equality of the temperatures and heat flows of both sides of the boundary should be fulfilled in order to identify the thermal interaction at the boundaries of two media. In the general case, the condition for conjugation can be written as:

$$k_1\left(\frac{\partial T_1}{\partial n}\right) = k_2\left(\frac{\partial T_2}{\partial n}\right) + q_g(x, y, z, t); \; T_1(x, y, z, t) = T_2(x, y, z, t) \quad (10)$$

where $q_g(x, y, z, t)$ is the source of heat at the surface of the border, and

T_1, T_2, k_1, k_2 are the temperatures and thermal conductivity coefficients of bordering media, respectively.

When there is no boundary between the media, the processes of release and absorption of heat and the condition for conjugation Eq. (10) acquires the following form:

$$k_1\left(\frac{\partial T_1}{\partial n}\right) = k_2\left(\frac{\partial T_2}{\partial n}\right); T_1(x, y, z, t) = T_2(x, y, z, t). \quad (11)$$

The given equation affects the heat exchange in the bordering areas.

The spatial extent of heat transfer is described by a time-dependant quantity that is called thermal depth of penetration z_T:

$$z_T(t) = \sqrt{4a_t t}, \quad (12)$$

which in turn means that the distance in which the temperature decreases is *e* times from its maximal size.

With reference to the thermal decomposition of tissues, it is important to adjust the duration of the laser pulse to reduce the thermal damage of the surrounding tissues. To this end, a parameter called thermal relaxation time is introduced. This parameter can be obtained if in Eq. (12) z_T is made equal to the optical penetration depth δ, hence

$$\delta = \sqrt{4a_t \tau_T}, \quad (13)$$

where τ_T is the thermal relaxation time.

Regarding the thermal decomposition, this parameter is very important because it measures the thermal susceptibility of tissues.

When the duration of the laser pulse is $\tau < \tau_T$, heat does not even penetrate at a distance equal to the optical penetration depth; subsequently, the thermal damage of the surrounding tissues is insignificantly reduced.

Nevertheless, if $\tau > \tau_T$, heat can diffuse to a distance much bigger than δ and the surrounding tissues can be highly damaged.

The Monte-Carlo method is suitable for modeling multiple laser beam scattering processes, in a multilayer environment [16]. For this purpose, the laser beam is modeled in different environments. The intensity of light on the boundary of the air and the bio layer increases up to 5-20% and sharply decreases in depth. The border conditions $T(z, 0) = T_0$, $z > 0$, Δ are a Laplace operator [3, 12] in cylindrical coordinates.

Study and modelling of lasers in thermal effects

When bio-tissues absorb laser radiation, heat is generated. This heat causes wide range of effects which correspond to different power levels that are injected and also leads to tissue heating within the physiological temperature, which in turn leads to reversible or irreversible changes in the state of the tissues.

The generated heat, described as a heat source S at point **r** which is proportional to the light flow intensity $\varphi(\mathbf{r})$ ([mW/cm^2]) and absorption coefficient $\mu_a(\mathbf{r})$ at this moment:

$$S(\mathbf{r}) = \mu_a(\mathbf{r})\varphi(\mathbf{r}). \tag{14}$$

The bioheat equation is written based on the energy balance in the system and describes the temporary change in bio-tissue temperature at a point from the tissues **r**:

$$\rho c \frac{\partial T(\mathbf{r},t)}{\partial t} = \nabla[k_m \nabla T(\mathbf{r},t)] + S(\mathbf{r}) + \rho c \omega (T_a - T_v), \tag{15}$$

where ρ is tissue density,

c – the specific temperature of the tissue,

$T(\mathbf{r},t)$ – the tissue temperature at time t,

k_m – coefficient of thermal conductivity,

$S(\mathbf{r})$ – the heat source term, which determines the speed of heat dissipation at point $\mathbf{r}$,

ω – the tissue perfusion rate,

T_a – the inlet arterial temperature,

T_v – the outlet venous temperature and all at point $\mathbf{r}$ in the tissue.

The equation does not take into consideration certain processes such as convection, reflection, vaporization, metabolic heat effects because they are negligible in many practical cases. It is assumed that the source term does not change during heating. The first term, to the right of the equal sign, describes heat conductivity (usually from point $\mathbf{r}$), and the source term accounts for the heat generation at the expense of the photons absorbed.

When laser beam interacts with tissues, the heat transfer by blood (last term of the equation) is negligibly small. Initial and boundary conditions must be taken into account in order for the equation to be solved.

The initial conditions are the tissue temperature at moment of time $t = 0$, and the boundary conditions depending on the tissue structure and the geometry of laser heating area.

Fig. 2 shows the modelling of temperature distribution in tissues in the implementation of the bioheat Eq. (15) based on the energy balance in the system, and describing the temporal change in temperature of the bio-tissues at a given point of space.

The study has been conducted in order for a temperature of 60 °C, 45 °C and 40 °C to be reached (Fig. 2) all for the same time duration of the treatment.

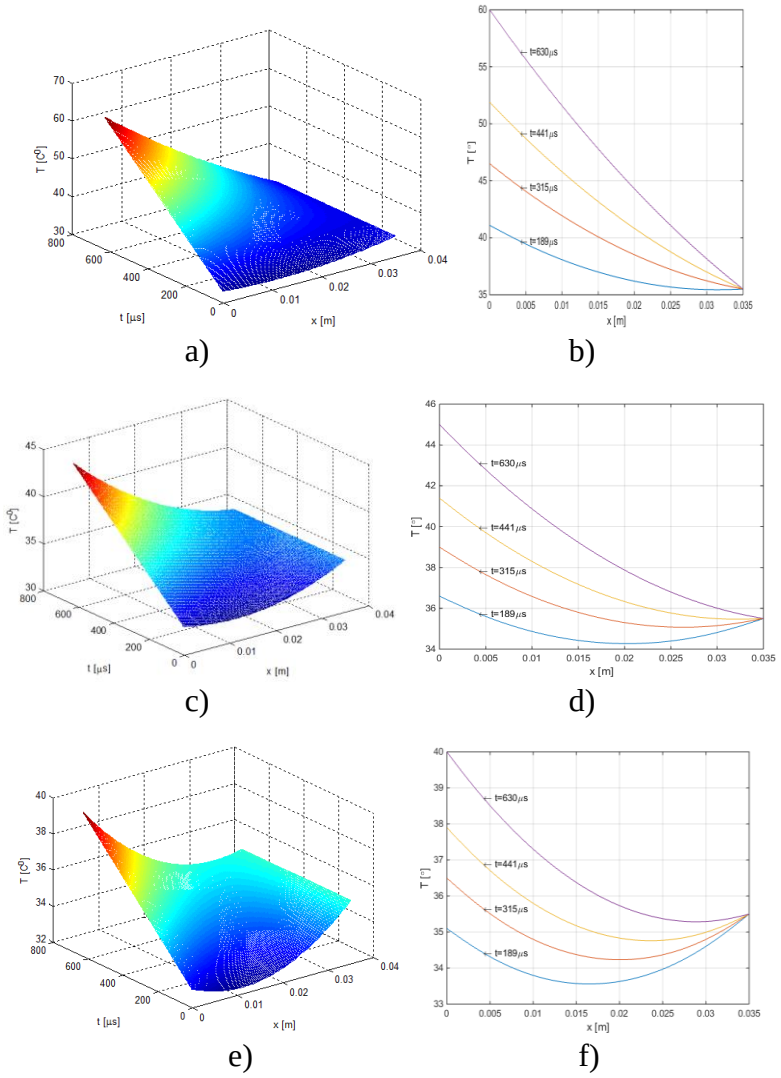

Fig. 2 Modelling of the distribution of temperature in tissues

Figs. 2a, 2c, and 2e show the dependence of the temperature distribution in the tissues depending on the penetration depth and the time of exposure. With the increase of the time of exposure, the temperature in the tissue expectedly increases. At the same time, there is non-linearity of the temperature distribution in depth of penetration into the tissue. Figs. 2b, 2d, 2f illustrate the dependence of the temperature distribution on the depth of tissue penetration for four different times of exposure. In this case, the desired temperature to reach is 60 °C, the non-linear character of the temperature distribution in the depth of the tissue is less pronounced. The dependence of the temperature distribution on the depth in the tissue is monotonic, the temperature decreasing with increasing depth. In case the desired temperature to reach is 45 °C, more pronounced non-linear character of the temperature distribution in the depth of the tissue is observed. In times of exposure of 441 μs and 630 μs, the dependence of the temperature on the penetration depth is monotonous, while in times of exposure of 189 μs and 315 μs, a minimum of the temperature at certain penetration depths is observed. The minima of the temperatures for the two times of exposure are observed at different depths in the tissue. In case the desired temperature to reach is 40 °C, a strong non-linear character of the temperature distribution in the depth of the tissue is observed. At all times of exposure, a minimum of the temperature is observed at certain penetration depths. The minima of the

temperatures for different times of exposure are observed at different depths in the tissue.

Conclusion

Modeling presents the opportunity for the calculation of the penetrating depth and the temperature of the medium on a biomaterial, which can be effectively used when choosing the individual dose of radiation.

The model could be used for numerical experiments, which allow more detailed investigations of the influence of various laser parameters (power density, exposure time, working regime, and wavelength) on the tissues and can demonstrate the extent of exposure on the wavelength. Which in turn allows for a reduction of the thermal exposure over the surrounding tissues resulting in a considerable decrease in biological decomposition of the tissue structure.

When the temperature raises to about 40 °C blood flow increases both in the tumor and in the healthy tissues. When the temperature reaches 41.5 °C it becomes toxic for the tumor cells, and temperature higher than 42.5 °C begins to damage the tumor tissue. The temperature effect increases sharply when it reaches 43 °C and the extent of cell death doubles at every 1 °C with which the temperature exceeds 43 °C. It has been proven that tumor tissues are much more sensitive to temperature changes than healthy tissues. A significant influence for this has the higher acidity (lower pH values) of tumor tissues due to the reduced oxygenation.

By using natural heat absorbers (melanin in the melanoma) or exogenous sensitizers, laser phototherapy can provide high levels of local temperature rise in the tumor without heating the surrounding healthy tissues. This leads to direct destruction of cancer cells and the release of large amounts of tumor antigens which contributes to the formation of anti-tumor immunity.

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