

## MODELING OF TEMPERATURE DISTRIBUTION IN BIOTISSUES

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### Abstract

One of the widely used ways to model the spread of laser beams in biological tissues is through the Monte Carlo method. This article focuses on the distribution of temperature in human skin based on the results of the Monte Carlo method for a laser beam with a wavelength of 800 *nm*. A mathematical model of the heat transfer process, based on the solution of the differential equations through the finite element method, was introduced. Simulations of the laser beam diffusion and the distribution of temperature in the skin were performed for two types of beams. The results of this work can be used to make an assessment of the energy levels of pulse laser treatment with different durations, during which a damage of biological tissue is observed, as well as to create standards for the safe use of laser systems.

**Keywords:** biological tissues, Monte-Carlo method, laser.

### 1. INTRODUCTION

There is a trend in modern biotechnology and physiology related to experimental and theoretical studies of heat transfer processes in the body as well as in its individual organs and tissues. Biological tissue is a specific physical environment with a complex structure and the study of heat transfer processes is essential.

The laws governing the laser beam penetration into tissues have a direct relation to the problem of the mechanism of biological impact of the laser beam. One of the reasons why the laser beam penetrates to a limited depth in tissues is because of the absorption of the beam and this is the initial link that precedes the sequence of changes that occur as a result of the organism being irradiated. The depth of the laser beam penetration into the biotissue is very important in practical terms, as it is one of the key factors for determining the possible limit of laser application.

Determining the mechanisms and the precise threshold characteristics of the field of ultrashort optical laser pulses in their effect on biological entities is of considerable importance for many practical and scientific applications in biotechnology and in medicine (Akhmanov, Vysloukh, Chirkin, 1988, 312 p.; Rulliere, 2005, 428 p.).

The mathematical modeling of the internal radiation of human biological tissues, based on the numerical solution of the thermal conduction equation, is of great practical significance (Antonov., Hristozov, 2017, pp. 310-325; Vesnin, Sedankin, 2012a, pp. 43–61; Vesnin, Kaplan, 2008b, pp. 82–87).

The work is aimed at solving the main problem of medicine, concerning the improvement of the efficiency of the methods for early treatment of oncological diseases (Losev, Mazepa, Zamechnik, 2014, p. 254). The conducted computational experiments will allow to investigate the influence of uncertainties in the spatial distribution of the physical parameters of biotissues on the temperature fields.

The work also examines the role of a complex small-scale structure in the formation of temperature distribution due to the strong spatial heterogeneity of the thermal conductivity of the various biological components. The transition from multilayer models to multicomponent biotissue, taking into account its structure and inhomogeneous parameters (thermal conductivity, water content, specific heat release, electrical conductivity, dielectric constant, thermal power) can improve the efficiency of thermal therapy of tumors at early stages (Jacques, 2011b, pp. 109–144).

The damage of biotissues is due to the temperature impact when laser optical radiation is absorbed, which leads to reversible or irreversible changes in its structure (Vogel, 2003, pp. 577–644; Wright, Barrett, Welch, 2002a). Irreversible damages can be divided into the following types, depending on the tissue temperature ( $T_{crit}$ ): coagulation  $T_{crit} = 45–100^{\circ}C$ , vaporization  $T_{crit} \geq 100^{\circ}C$ . At a temperature of  $100^{\circ}C$ , the water molecules contained in most biotissues begin to evaporate. The large thermal vaporization of water is useful as far as it helps to remove excess heat and prevents overheating of adjacent tissues. If the water molecules evaporate and the laser effect continues, a further increase in tissue temperature will follow. After the water evaporates, the dehydrated tissues are very quickly heated to a temperature of  $150^{\circ}C$ , which initiates the process of carbonization of the tissues (Müller, Sliney, 1989, p.253; Sliney, Trokel, 1993; Tuchin, 2016a, p. 030201; Welch, 2011a).

## 2. DETERMINING THE DEPTH OF PENETRATION OF OPTICAL RADIATION BY USING THE MONTE-CARLO METHOD.

The numerical modulation of the diffusion of laser radiation with wavelength of  $800\text{ nm}$  in biotissues, using the Monte-Carlo method (Jacques, 2011b, pp. 109–144; Meglinski, Doronin, 2012) allows us to obtain data on the intensity distribution in the penetration depth.

In the mathematical modeling, the Henyey-Greenstein function is accepted as the scattering phase function:

$$p(\theta) = \frac{1 - g^2}{(1 + g^2 - 2g \cos\theta)^{3/2}},$$

where  $p(\theta)$  is a probability function,  $\theta$  is a scattering angle,  $g$  is the average cosine of the scattering angle. For the  $\theta$  angle we obtain:

$$\theta = \arccos \left( \frac{1 + g^2 - \left( \frac{1 - g^2}{1 + g^2 - 2gR} \right)^2}{2g} \right),$$

where  $R$  is a random evenly distributed number in the interval  $(0,1)$ .

The experiment was performed with a beam with a Gaussian distribution of  $0,1\text{ cm}$  and focused Gaussian beam with a pulse duration of  $500\text{ ns}$  and wavelength of  $800\text{ nm}$ , refractive index  $n = 1,4$  and mean cosine of the scattering angle  $g$ . Optical parameters of human tissues are measured in *in vitro* conditions and are presented in Table 1 (Tuchin, Utz, Yaroslavsky, 1994, pp. 3178–3188; Tuchin, Utz, Yaroslavsky, 1993a, pp. 234–258).

**Table 1. Optical parameters of biotissues**

Biotissue	$\lambda, \text{nm}$	$\mu_a, \text{cm}^{-1}$	$\mu_s, \text{cm}^{-1}$	$g$
Dermis	800	1,7	175	0,85

Figure 1 shows the results from the Monte-Carlo simulated photon distribution in the dermis for a beam with Gaussian distribution and radius of  $5 \times 10^{-3} \text{ cm}$  and a focused Gaussian beam with the same radius according to the penetration depth (on the z axis) and the radius of remoteness from the z axis.

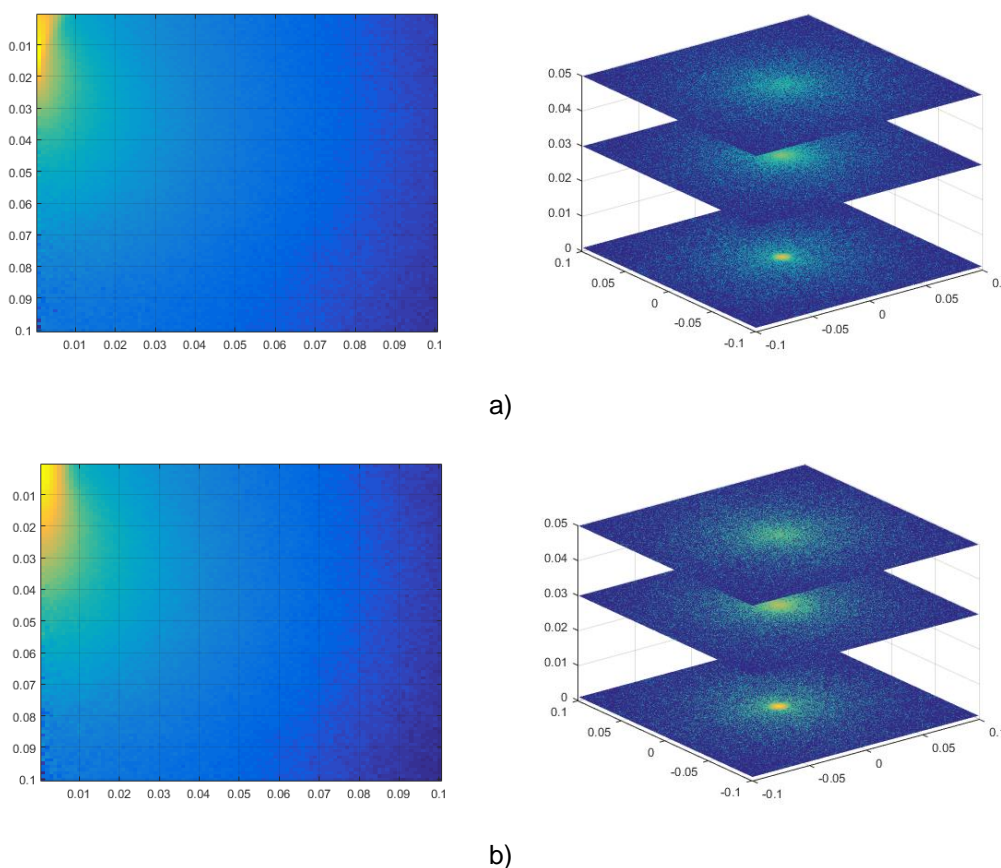


Figure 1. Density distribution of the photon transfer in dermis of: a) a focused Gaussian beam and b) a beam with a Gaussian distribution.

The obtained results show that the scattering of a focussed Gaussian beam in the dermis is less than the scattering of a beam with Gaussian distribution. Figure 1a clearly illustrates that the beam initially focuses upon penetration in the dermis and then remains with a relatively constant width that is smaller than that of the non-focussed beam at the same depth of penetration (Figure 1b). At depths greater than  $0.035 \text{ cm}$ , both beams diminish considerably, as this is more pronounced for the focused beam.

The nature of the interaction of laser radiation with biological tissues depends on the absorption coefficient for a particular wavelength. The examined modified Monte Carlo method has several advantages. This method is applicable to environments with different geometries and it produces three-dimensional information about the beam diffusion in tissue. When emitting 1 million photons, the error of the method does not exceed 1% of the value obtained.

### 3. MODELING OF HEAT PROCESSES.

When biotissues absorb laser radiation, heat is generated. The modeling should be solving the problem for optimization of the laser emitter parameters and should evaluate the results obtained from the laser effect of the already selected laser on the biological environment. The generated heat causes a number of effects, which correspond to different power levels that are used, and 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 source of heat  $S$  at point  $\mathbf{r}$ , which is proportional to the fluence rate of light  $\varphi(\mathbf{r})$  ( $mW/cm^2$ ) and absorption coefficient  $\mu_a(\mathbf{r})$  at this point:

$$S(\mathbf{r}) = \mu_a(\mathbf{r})\varphi(\mathbf{r}). \quad (1)$$

The bioheat equation is written based on the energy balance in the system and describes the temporary

change in biotissue temperature at a given point in space  $\mathbf{r}$ , in the absence of blood flow in the tissue:

$$\rho c \frac{\partial T(\mathbf{r}, t)}{\partial t} = \nabla [k_m \nabla T(\mathbf{r}, t)] + S(\mathbf{r}), \quad (2)$$

where  $\rho$  is the tissue density,  $c$  is the tissue-specific heat,  $T(\mathbf{r}, t)$  is the tissue temperature at time  $t$ ,  $k_m$  is the coefficient of thermal conductivity,  $S(\mathbf{r})$  term that describes the heat source which determines the speed of heat release at point  $\mathbf{r}$ .

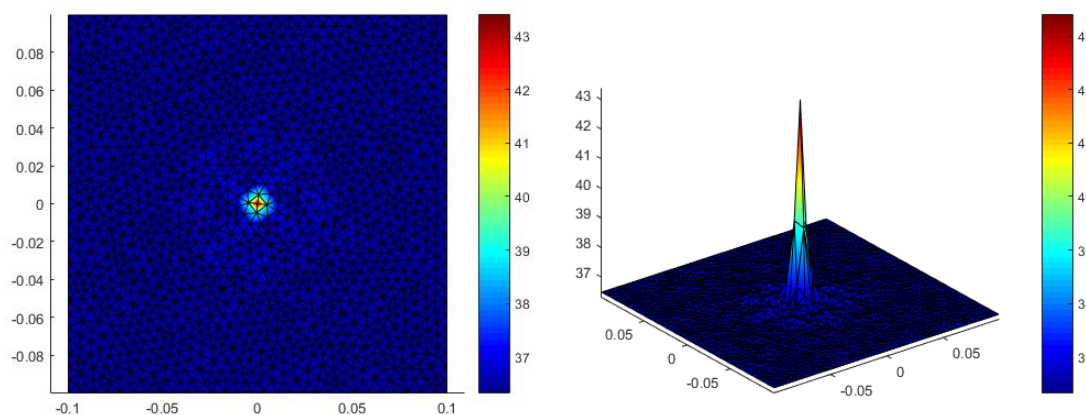
In this equation certain things, such as convection, reflection, vaporization, metabolic heat effects are not taken into consideration as they are negligible in many practical cases. It is assumed that the source term is constant over the time interval of heating. The first term, to the right of the equal sign, describes the heat conduction (usually from point  $\mathbf{r}$ ), and the source term accounts heat generation at the expense of photon absorption. Initial and boundary conditions must be taken into account in order to solve this equation. The initial conditions are the tissue temperature at moment of time  $t = 0$ , and the boundary conditions depend on the tissue structure and the geometry of laser heating (Friedman, 2011a, pp. 1189–1198).

Figure 2 shows the results of the modeling of temperature distribution in tissues during the implementation of the bioheat equation (2) on the basis of the energy balance in the system, the solution being obtained by applying the finite element method. The study was performed for the biological structure of the dermis. The data used for the dermis is shown in Table 2 (Tuchin, Utz, Yaroslavsky, 1994, pp. 3178-3188; Tuchin, Utz, Yaroslavsky, 1993a, pp. 234–258).

**Table 2. Biological structure of the dermis**

Biotissue	$\rho, [kg/m^3]$	$c, [J/kg.K]$	$k, [W/m.K]$
Dermis	1200	3800	0,53

Figure 2 shows the temperature distribution, as focused Gaussian beam and a beam with Gaussian distribution were investigated at the same duration of impact of the beam.



a)

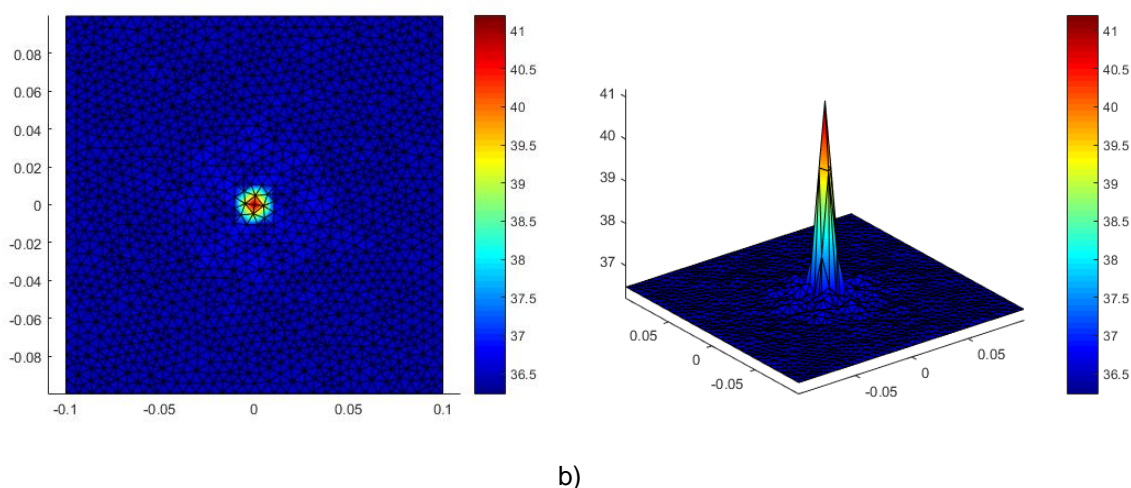


Figure 2. Temperature distribution (a-b).

From Figure 2a, it can be observed that the focussed Gaussian beam heats the tissue to a temperature of  $43^{\circ}\text{C}$ . And Figure 2b shows a distribution of a beam with Gaussian distribution temperature at which the tissue temperature reaches  $41^{\circ}\text{C}$ . The figure shows that the Gaussian beam temperature distribution covers a larger area of the section to which the beam has reached, compared to the focused beam. This indicates that the use of a focused beam is more efficient and allows for more precise heating of only the desired part of the tissue.

#### 4. DISCUSSION.

The morphological diversity of tissues determines the different nature of light passing through them. The size of the cells, of the cellular structural elements and of the connective tissues is within the range from a few tens of centimeters to hundreds of micrometers (Bagratashvili, Sobol, Shechter, 2006; Saratov, 2000; Tuchina, 2007; Tuchin, 2011). It should be noted that when laser radiation is applied to the skin, damage to the biological tissues occurs, which cannot be explained by the standard thermal melting mechanism (Fredriksson, 2008a). The difference is explained by the multiphoton excitation of electrons, which occurs at a time equal to the duration of the pulse when the energy is transferred directly to the atomic subsystem (Dyukin, Martsinovskii, Shandybina, Yakovlev, Nikiforov, Guk, 2011, pp. 558–562; Stampfli, 1990, pp. 7163–7173). In this case, the water molecules serve as a transmission link which absorbs energy from the laser radiation and excites the vibrational modes of collagen with a relaxation time of about 3 ps. (Kropman, Bakker, 2001, pp. 2118–2120; Kropman, Nienhuys, Bakker, 2002, p. 077601). At various temperature ranges, the biological tissue can undergo different chemical and structural changes. At  $T = 40 \div 45^{\circ}\text{C}$  irreversible damage to the tissues is not to be expected (enzymes may be activated and changes in the membrane or cell death may occur only with sufficiently long heating). Heating of tissues to a temperature of  $42 \div 45^{\circ}\text{C}$  degrees in medical practice is used for local thermotherapy (Parvanov, Sotirov, Sakakushev, 2006a, Sakakushev, Sotirov, Parvanov, 2006b; Sotirov, Sakakushev, Koriykov, Georgiev, Zhel, Popesku, 2010; Sotirov, Sakakushev, Tonev, Georgiev, 2006c; Sotirov, Ivanov, Sakakushev, 2016b; Parvanov, Sakakushev, Sotirov, 2014a; Sakakushev, Sotirov, Parvanov, Angelov, 2014b).

#### 5. CONCLUSIONS.

The wavelength and the laser radiation power are determined by the size of the tumor and the absorption spectrum of the pathological tissue. All types of laser thermotherapy require careful determination of the laser radiation dosimetry, reliable data on the optical and thermophysical parameters of the tissues and the use of the means to control these parameters. The developed mathematical model describes the heat transfer processes based on an analytical solution of a system of equations describing the dynamics of the electron and atomic subsystems, the results of the experiment showing that the scattering of a focused Gaussian beam in the dermis is less than that of a beam with Gaussian distribution. It can clearly be seen how the beam initially focuses upon penetration in the dermis and then remains with a relatively constant width that is smaller than that of the non-focussed beam at the same depth of penetration. At depths greater than  $0.035\text{ cm}$ , both beams diminish considerably, and this is more pronounced for the focused beam. The beam with Gaussian distribution covers a larger area of the section to which the beam has reached

compared to the focused beam. This indicates that the use of a focused beam is more efficient and allows for more precise heating of the desired part of the tissue (Petrova, 2016; Petrova, Petrov, 2018, pp.213-228; Petrova, Petrov, 2014a, pp. 201-206; Tuchin, 2007a).

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