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HEAT FLOW IN BIOLOGICAL TISSUE UNDERGOING LASER IRRADIATION

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Abstract. In this paper we present a numerical method to determine thermal effects in biological tissue undergoing laser irradiation. The scattering and absorption of the laser beam have been analysed. Results are compared with experimental observations.

INTRODUCTION

Medical applications of lasers have focused particular interest on the interaction of infra red laser light in biological tissue. High intensity light sources are under continual investigation because of the importance of the problem in medicine. Lasers are very convenient tools by which the problem can be investigated. For several years, lasers have been used to operate damaged retinae, thus preventing blindness. More recently, lasers have been used by surgeons to cut into tissues (carbon dioxide laser scalpel), to stop internal hemorrhage (neodymium: YAG and argon laser photocoagulation) and to treat tumors. There exist several types of lasers used in medicine; the most often met are Argon, Nd: YAG, runbin and CO_2. Although the clinical applications of lasers are rapidly multiplying, it still is not clear how the laser exerts its effets. The physical heterogeneity of biological tissue results in considerable region variation in light absorption and scattering, two of the critical processes that determine the pattern of heating. The effects of this heterogeneity are poorly understood. Futhermore, the effects of multiple

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scattering of the incident laser beam, which play a dominant role in the case of the neodymium: YAG laser, have not been properly taken into account in previous works. The thermal effects in laser irradiation have here, however, a prime significance (Auth,1981; Anderson et al., 1981 ; Bellina and Seto, 1980; Halldorsson and Langerholc,1978 ; Halldorsson et al.,1981; Hofstetter and Frank, 1980; Kiefhaber et al., 1977). In this study we have developed a finite element method to consider the classic problem of light scattering, light absorption and thermal diffusion, which allows for spatial and temperature dependent variation of these events. Temperature-dependent variables are related to the physical changes that tissues undergo during heating, including coagulation, carbonization and vaporization. The computer analysis will be made using the finite element program carried out for ODRA 1305 computer (Służalec, 1984,1985). The results obtained may be very useful in studies under applications of lasers in medicine.

HEAT FLOW WITHIN THE TISSUE UNDERGOING LASER IRRADIATION

Heat flow equation

In this paper we will analyse only a two-dimensional heat conduction equation. Such a simplification is used in three-dimensional axisymmetrical problems.

The variation of temperature θ with time t in a two-dimensional region Ω , relative to the Cartesian coordinates (r,z), is governed by the equation

$$\rho c \frac{\partial \theta}{\partial t} = \underset{\sim}{\nabla}(k \underset{\sim}{\nabla} \theta) + Q, \quad \underset{\sim}{\nabla} = \left[\frac{\partial}{\partial r}, \frac{\partial}{\partial z}\right], \quad (r,z) \in \Omega, \tag{1}$$

where k is the temperature dependent conductivity , c is the temperature dependent heat capacity, ρ is the density, and Q is the rate of heat generation.

At the surface of body the temperature may be prescribed or the flow of heat due to convection may be specified or a combination of these conditions may exist. The region Ω is divided into a number of eight noded isoparametric elements Ω^e , with quadratic shape function N^i associated with each node i . The unknown function θ is approximated through the solution domain at any time t by

$$\theta = \sum_{i=1}^{n} N^i \theta^i(t) = \underset{\sim}{N}\,\underset{\sim}{\theta} \,. \tag{2}$$

where $\underset{\sim}{\theta}$ is the column vector of nodal values θ^i . The substitution of expansion (2) into equation (1) and the application of the Galerkin method produce the following equation

$$\underset{\approx}{C}\,\dot{\underset{\sim}{\theta}} + \underset{\approx}{K}\,\underset{\sim}{\theta} + \underset{\sim}{F} = \underset{\sim}{0} \,. \tag{3}$$

The form of the matrices $\underset{\approx}{C}$, $\underset{\approx}{K}$ and $\underset{\sim}{F}$, together with a description of the temporal discretization of equation (3) and the resulting method of solution of the subsequent equations have been described by many authors (e.g. Comini et al., 1974; Morgan et al.,1978; Służalec, 1984,1985) and will not be considered further.

Modelling of the phase change process

At this point we present general information on computer analysis of heat flow with phase change. The readers interested in further details of this problem are referred to Comini et al.,(1974), Morgan et al.,(1978) or Służalec(1984,1985).

In constructing the solution of a problem involving phase change, a possible approach could be to track accurately the position of phase boundary and then to solve equation (1). In this paper the phase change process is modelled by a variant of the enthalpy method. In this method the phase change is assumed to occur over a temperature range and the associated latent heat effect is handled by increasing suitably the specific heat in this range. Thus if the phase change assumed to occur over the temperature interval $[\theta_L, \theta_V]$ where θ_L is the liquidus temperature, and θ_V is the vaporization temperature, then the specific heat c_Δ used in the calculation is defined by

$$c_\Delta = c + [H(\theta - \theta_L) - H(\theta - \theta_V)] L/\Delta\theta \tag{4}$$

where H denotes the Heaviside function, L is the latent heat and $\Delta\theta = \theta_V - \theta_L$ is the phase change interval. When this method is applied to the analysis of pure materials, in which the phase change occurs at a specific temperature (i.e. $\theta_L = \theta_V$) , a phase change interval $\Delta\theta$ $(\neq 0)$ must be assumed.

It has been demonstrated by Comini et al. (1974) and Morgan et al., (1978) that reasonable results can be obtained for the problems involving conduction, provided that a right choice is made for the value of $\Delta\theta$. For materials, however, in which the phase change does indeed occur over a reasonable temperature range this problem does not arise and the actual physical values of θ_L and θ_V can then be used successfully. In our analysis we will consider the tissue as a homogeneous structure of thermal properties like water. Such an approach is usually met in macroanalysis (see for instance Służalec and Muskalski, 1985 ; Kovtun et al., 1980; Hofstetter and Frank, 1980) .

Absorption and scattering problems

Here we present absorption and scattering problems appearing in biological tissue under the influence of the laser irradiation. Let $I(\underset{\sim}{r}, \underset{\sim}{\hat{n}})$ be the total intensity of radiation along the direction represented by the unit vector $\underset{\sim}{\hat{n}}$ at the position $\underset{\sim}{r}$ inside the tissue, which is produced by a laser beam incident on the surface of the tissue. The change in the intensity due to the combined effect of absorption and scattering is determined by the equation, which tekes the form (Mihalas, 1980)

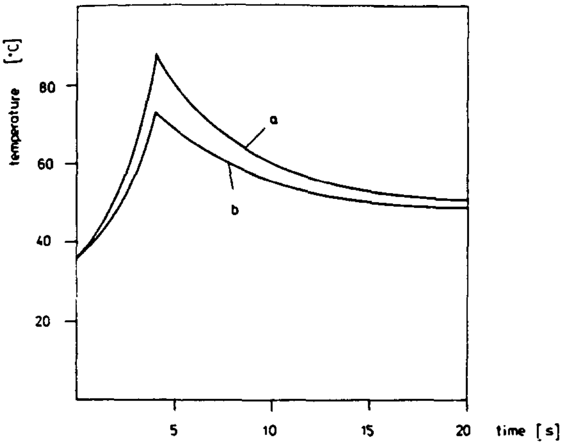
$$\Delta I(\underset{\sim}{r},\underset{\sim}{\hat{n}}) = (\alpha+\beta)\, I(\underset{\sim}{r},\underset{\sim}{\hat{n}}) + \beta/4(\underset{\sim}{r}) \int d\Omega'\, I(\underset{\sim}{r},\underset{\sim}{\hat{n}}'). \tag{5}$$

The integral is over the solid angle subtended by $\underset{\sim}{\hat{n}}'$, and α and β are the absorption and scattering coefficients, respectively, that may depend on $\underset{\sim}{r}$.

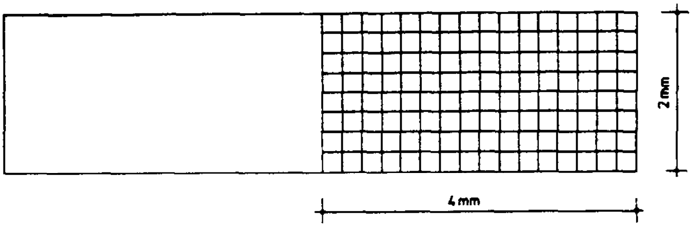
We will approximate the solution of equation (5) by replacing the integral over solid angle by a sum over finite directions n, and by introducing these relations to nodes of the finite element mesh. For the case of cylindrical symmetry, the simplest approximation is to include only the radiation $I(\underset{\sim}{r},\pm\underset{\sim}{\hat{z}})$ along and opposite the laser beam direction, which has been chosen as the z-axis, and the radiation $I(\underset{\sim}{r},\pm\underset{\sim}{\hat{r}})$ in the directions perpendicular to this axis. Then equation (5) may be written as

$$I^{k,s^+}_{+\hat{z}} = I^{k}_{+\hat{z}} - (\alpha+\beta)\, d(k,s^+)\, I^{k}_{+\hat{z}} + \beta/4\, d(k,s^+)\, I^{k},$$
$$I^{k,s^-}_{-\hat{z}} = I^{k}_{-\hat{z}} - (\alpha+\beta)\, d(k,s^-)\, I^{k}_{-\hat{z}} + \beta/4\, d(k,s^-)\, I^{k},$$
$$I^{k^+,s}_{+\hat{r}} = I^{s}_{+\hat{r}} - (\alpha+\beta)\, d(k^+,s)\, I^{s}_{+\hat{r}} + \beta/4\, d(k^+,s)\, I^{k},$$
$$I^{k^-,s}_{-\hat{r}} = I^{s}_{-\hat{r}} - (\alpha+\beta)\, d(k^-,s)\, I^{s}_{-\hat{r}} + \beta/4\, d(k^-,s)\, I^{k},$$

where $I^k = I^k_{+\hat{z}} + I^k_{-\hat{z}} + I^k_{+\hat{r}} + I^k_{-\hat{r}}$ is the total radiation intensity at node k, $I^{k,s^+}_{+\hat{z}}$ is the intensity at node s^+ abutting to node k computed in the direction of the axis z, $I^{k^+,s^-}_{-\hat{z}}$ is the intensity at nodes s^- abutting to node k komputed in the opposite direction to the axis z, $I^{k^+,s}_{+\hat{r}}$ is the intensity at node k^+ abutting to node s computed in the direction of the axis r, and $I^{k^-,s}_{-\hat{r}}$ is the intensity at node k^- abutting to node s computed in the opposite direction to the axis r, d (k,s) is the distance between nodes k and s.



·Fig.1 Temperature (a) on the front surface , (b) on the rear surface of the specimen



'Fig.2 Finite element model of the tissue

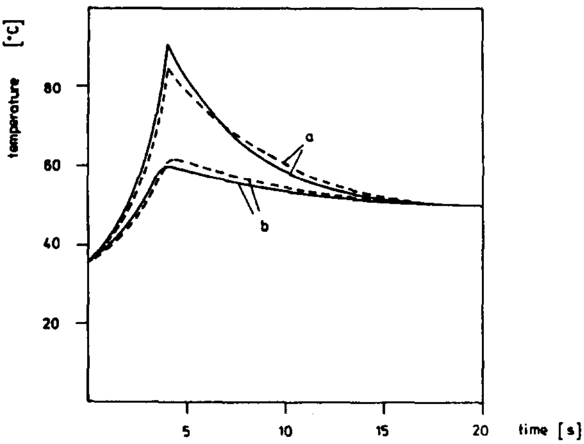


Fig.3 Influence of the coefficient α on the value of temperature in the tissue for β = 10, (a) $\alpha = 8\times10^{-2}$, (b) $\beta = 6.5\times10^{-3}$

The discretization used here is very simple from the mathematical point of view, because equation (5) is transformed to describe the intensity of radiation only in the nodal points of the finite element mesh.

NUMERICAL ANALYSES

Based on the theoretical formulations described above we present several results showing the temperature field in biological tissue under the influence of laser radiation. In the work of Halldorsson et al., (1981) the temperature measurements in bloodless dog have been made in response to the Nd: YAG laser radiation. Figure 1 compares the theoretical and observed temperatures as a function of time when a 2mm thick section of canine stomach is irradiated using a power setting of 50 W distributed over a circular area of 3mm diameter. Finite element model of the tissue is presented in Fig. 2. The time duration of pulse is 4 s. Both front and back surface temperature of the tissue are measured. The initial temperature = $36^{o}C$, environmental temperature = $30^{o}C$. On the surface of the tissue convection boundary conditions are assumed. Full lines denote theoretical temperatures, and dotted lines, measured. The following material properties have been used for analysis: $\alpha = 0.085$, $\beta = 10$, $c = 4.19 \times 10^{3} J kg^{-1} C$, $\rho = 10^{3} kg m^{-3}$, $k = 0.599 J m s^{-1} {}^{\circ}C$, surface film conductance 0.25 $W m^{-2}$. From the works of Anderson and Parnish, (1981); Halldorsson et al., (1981); Hofstetter and Frank, (1980); Johnston et al. (1980); Kiefhaber et al., (1977), it implies that the values of absorption and scattering coefficients should be determined experimentally for different kinds of tissue, because they vary considerably. To the numerical analysis these coefficients have been assumed to be like in the works of Anderson and Parnish (1981),

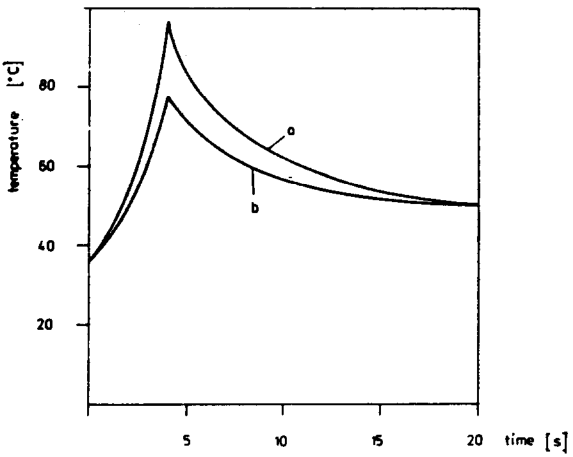


Fig.4 Influence of the coefficient β on the temperature distribution the tissue for $\alpha = 8.5 \times 10^{-3}$ (a) $\beta = 13$, (b) $\beta = 3$

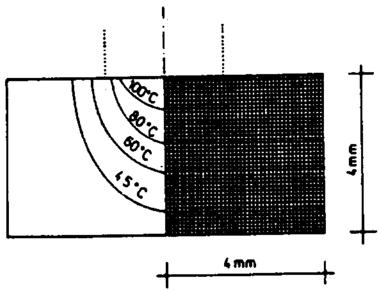


Fig.5 Model of the tissue and the obtained temperatures during heating with vaporization. Exposition time 1.7s

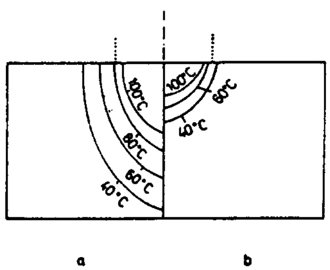


Fig.6 Temperature distribution in the tissue undergoing laser irradiation. Diameter of the beam 2.5 mm.

and Hofstetter and Frank (1980). The values chosen here were taken from existing measurements in the literature considering the situation similar to the presented in the paper. Figures 3 and 4 show the changes of temperatures in the tissue for different values of α and β. Figure 5 describes theoretical temperature distribution in the tissue assuming its initial temperature $36^{\circ}C$ and shows its finite element model. The assumed parameters of the process: power 40W, time 4s, laser beam diameter 3 mm, $\alpha = 8.5 \times 10^{-3}$, $\beta = 10$. From this figure we may determine the range of coagulation appearing at the temperature $60^{\circ}C$. We assume here that at $100^{\circ}C$ the vaporization process occurs. After vaporization the analysed tissue is treated as a void space. Since in the presented analysis we consider the nonlinear heat conduction equation there is no difficulty in introducing in the temperature range more than $100^{\circ}C$ special properties of the tissue and treating it as a void space. The latent heat is equal to 2×10^{3} kJ kg^{-1}. Figure 6 presents the computed results of the temperature distribution in biological tissue using the lasers Nd: YAG and argon. Figure 6 a is for the parameters 80 W, 1.5 s (as for Nd: YAG laser), Fig. 6b for 8W and 0.31 s (as for the argon laser).

FINAL REMARKS

In this paper we present a theoretical model to determine the temperatures in biological tissue under the influence of laser irradiation. We give the examples of analyses and compare them with experimental investigations. The model and the results so far are valid for tissue that is not perfused by blood. In medicine these applications are interesting but rather rare. The presented model could be expanded to the situation of heat flow in the vascular system. The problem is however very complicated. Such an analysis will be undertaken in subsequent publications. The method presented in the paper may be very useful in determination of parameters of laser irradiation. Such an application is also very important for medical surgery. Because the clinical applications of lasers are rapidly multiplying, the problems of thermal and thermo-mechanical effects in the tissue are very significant.

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