Journal of Engineering Physics and Thermophysics, Vol. 75, No. 4, 2002

HEATING OF INTRAOCULAR MEDIA BY NEAR IR LASER RADIATION

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UDC 621.378:612.085.1

The temperature fields in the refractive eye tissues at a different focusing of near IR laser radiation have been determined numerically. Depending on the focusing and the radiation wavelength, the levels and regions of localization of the maximum heating of intraocular media have been calculated. New prospects of using these lasers in ophthalmology have been contemplated and the necessity of additional studies of the conditions for their safe use is shown.

In using lasers in medicine, an important role is played by the proper, scientifically substantiated choice of tissue irradiation conditions. These conditions depend on a complex of factors, including the specificity of a pathology, the localization of the object being irradiated, the optophysical, mechanical, and thermophysical properties of tissues, etc. The choice of irradiation conditions requires, as a rule, serious experimental and theoretical investigations based on the studies of the main mechanisms of interaction between the laser radiation and biological tissues. In particular, the basis for the destructive action of powerful lasers in surgical operations is the optical radiation energy inversion to heat.

The present paper theoretically analyzes the dynamics of heating tissues by near-infrared lasers of the 1–2.4-μm range as applied to problems of intraocular microsurgery.

The spectral absorption coefficients of the majority of intraocular media are close to the absorption coefficients of water and vary over the range of wavelengths under consideration from a few to hundreds of inverse centimeters. One of the most promising directions of using IR laser radiation having large absorption coefficients is refraction ophthalmosurgery. Here near-infrared lasers can be an alternative to ultra-violet (UV) excimer lasers [1], since the probability of a mutagenic action of the IR radiation is a few orders of magnitude lower than of UV.

On the other hand, the radiation in the spectral ranges corresponding to small absorption coefficients (a few inverse centimeters) provides a number of new possibilities in laser intraocular surgical operations both in the bulk of the vitreous body (Fig. 1, position b) and in the rear pole (positions 7–13).

In particular, one of the topical problems of modern laser ophthalmology is the laser coagulation of pathological neoplasms in the vitreous body that are characteristic of sugar diabetics. The result of blood circulation disturbances at this pathology is ischemia (lack of blood supply) in the retina stimulating intergrowth of new blood vessels. Such neoplasms are more prone to subsequent hemorrhages in the vitreous body.

Simple diabetic retinopathy (DR) is characterized by the presence of punctate hemorrhages, punctate foci of solid inclusions, and nonuniform vasodilation. The second stage of DR is attended by pronounced changes in the fundus in the form of multiple hemorrhages, venous anomalies, trichangiestasis, and by the formation in the choroid of conjugates between arterioles and veins. At the third state of DR, in addition to the above-mentioned changes, the formation of new vessels not only throughout the retina field, but also in the area of the optic disk with penetration into the vitreous body is revealed. Repeated hemorrhages in the vitreous body lead to the formation of fibrous tissue and development of cords from the convective tissue followed by the possible retinal detachment and loss of vision. In the case where in the vitreous body a considerable amount of convective tissue is formed, the fundus at ophthalmoscopy is inaccessible for investigation.

The newly formed vessels, thin and fragile, are prone to repeated ruptures and lead to new hemorrhages. At this stage of retinopathy, there appear dimness foci and new formations of peripheral vessels of the retina and of vitreous body vessels, which cause retinal detachment and rupture and the development of secondary glaucoma.

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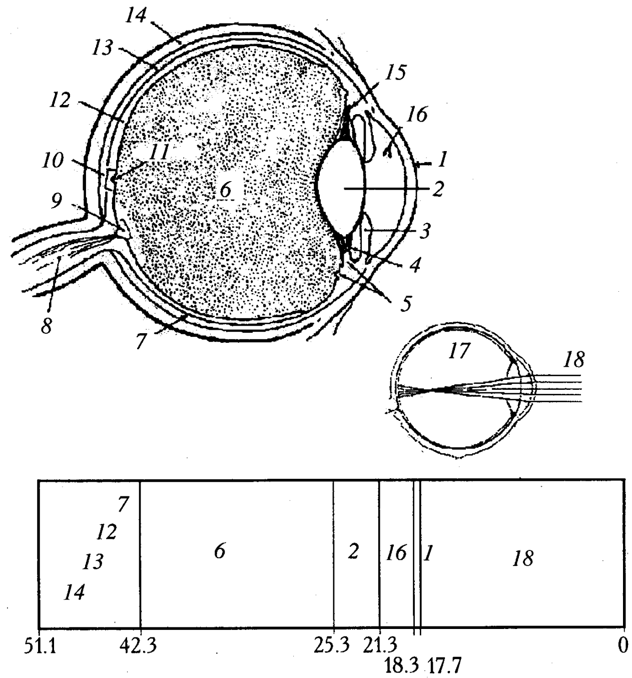

Fig. 1. Scheme of the anatomic structure of the eye and the ray path in the eye: 1) cornea; 2) crystalline lens; 3) iris; 4) trabecula; 5) cillary body; 6) vitreous body; 7) pigment epithelium; 8) optic nerve; 9) blind spot; 10) macula; 11) central fossa; 12) retina; 13) choroid; 14) sclera; 15) cylindrical muscle; 16) front chamber; 17) scheme of the ray path in the refractive media of the eye; 18) air layer adjacent to the cornea. The lower scheme shows the sizes of the estimated volume in millimeters.

The tasks of laser microsurgery are versatile. They include blocking of the blood flow in vessel neoplasms, destruction of anomalous structural elements (adhesion), coagulation of hemorrhage foci, and many other things. The common point in performing these tasks is the necessity of focusing the laser radiation into a given area inside the eye and provision of a strictly localized heating of the tissue to be destructed (coagulated) along with the most sparing action on adjacent healthy structures.

Below we consider the heating of intraocular media by collimated IR radiation having relatively low (of the order of a few inverse centimeters) spectral absorption coefficients. In this case, as the radiation propagates in the cornea-retina section, the sequential decrease in the power density associated with the absorption competes with its increase caused by the focusing. The spatial fluctuations of the temperature field formed thereby need to be studied in detail for both the possible use of such effects in the above-formulated task of practical ophthalmology and from the point of view of correct estimation of the danger of such a radiation to the eyes.

Determination of the tissue temperature is based on the numerical solution of the heat-conduction problem by a method that makes it possible (unlike the method considered in [2–5]) to take into account the laser-beam convergence along with its attenuation in a partially absorbing medium. The temperature field is determined by solving the nonstationary heat-conduction equation in a cylindrical area with axial symmetry. The condition of temperature conjugation at the boundary between the cornea surface and the surrounding air is formulated as the presence of a heat flow from the cornea to the air only due to the heat conduction of the air, in the absence of convection. For pulsed heating this assumption is valid, since the initial period of the development of natural convection [6] at the considered values of the irradiation power density constitutes a few tenths of a second.

The visible radiation is predominantly absorbed in the fundus tissues by the pigment epithelium of the retina (Fig. 1, 7, 12), whose main absorbing element is pigment melanoprotein granules. In the region of near IR radiation, it is also necessary to take into account the absorption in the other refractive media — the cornea, the chamber moisture, the crystalline lens, and the vitreous body. The measurements of the absorption coefficients in the near IR region

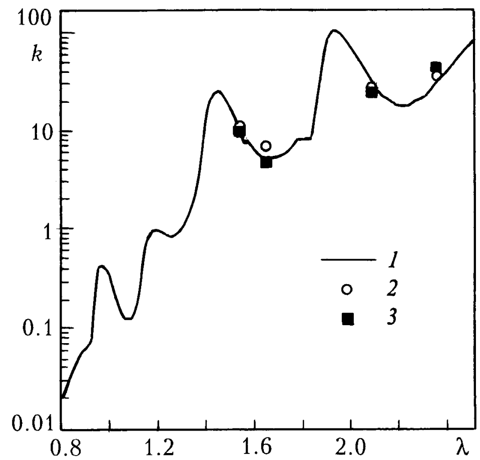

Fig. 2. Spectral dependence of the absorption coefficients of water and the tissues of the front section of the eye in the wavelength range from 0.8 to 2.5 μm: 1) water; 2) cornea; 3) front chamber. k, cm^{-1}; λ, μm.

for the cornea and the front chamber moisture made at the Biophysics Institute of the Health Ministry of the Russian Federation for four wavelengths [5] have shown that for these tissues (Fig. 2) the absorption coefficients are close to the absorption coefficients of water.

At exposures longer than 10^{-5} sec the tissue lesion is mainly of a thermal nature. However, it has been established [7] that prolonged (of the order of a few minutes) and systematic exposures of the eye tissues to laser radiations required to raise the temperature even a few fractions of a degree can lead to irreversible morphofunctional changes in them.

The mathematical model under consideration takes into account the difference in both the thermophysical properties of the tissue layers and their spectral absorption coefficients. The presence of the blood flow in the choroid capillaries is taken into account, as in [2, 8, 9], in determining the effective heat conductivity of tissues.

The initial temperature corresponds to the physiological conditions. The spectral absorption coefficients of the cornea, the front chamber, the crystalline lens, and the vitreous body are taken to be equal to the absorption coefficients of water, and the absorption coefficient for the fundus tissues (Fig. 1, 7, 12, 13) is assumed to be independent of the wavelength in the spectral region under consideration; it is of the order of 600 cm^{-1}. The heat-conduction equation written with allowance for these features is solved using the locally one-dimensional economic difference scheme.

Geometrically, the eye model under consideration represents a multilayer cylinder, and the sizes and optical and thermophysical properties of the layers correspond to the real structure of the eye (Fig. 1). The Z-axis of the cylinder coincides with the optical axis of the eye. The cylinder diameter is chosen to be much larger than the beam diameter so that within the estimated time the temperature field in the vicinity of the boundaries remains unperturbed. The radiation beam propagates along the cylinder axis and has in the plane r, Z an axially symmetric Gaussian radial intensity distribution. In the calculations given below, the cylinder radius is 16 mm. The estimated volume includes the air layer adjacent to the cornea, which is taken to be 17.7 mm, and an 8.8-mm-deep tissue layer. These are beyond the eye coat.

The numerical solution of the equation

$$\rho(Z)\,C_p(Z)\,\frac{\partial T}{\partial t}=\frac{1}{r}\frac{\partial}{\partial r}\left[r\kappa(Z)\,\frac{\partial T}{\partial r}\right]+\frac{\partial}{\partial Z}\left[\kappa(Z)\,\frac{\partial T}{\partial Z}\right]+Q(t)\,q(r, Z, k, s) \tag{1}$$

was carried out by breaking down the differential equation in partial derivatives into a chain of one-dimensional difference problems. The difference scheme has a second order of approximation and is completely stable. The boundary conditions for the temperature are of the following form, taking into account that it was counted off from the physiological value:

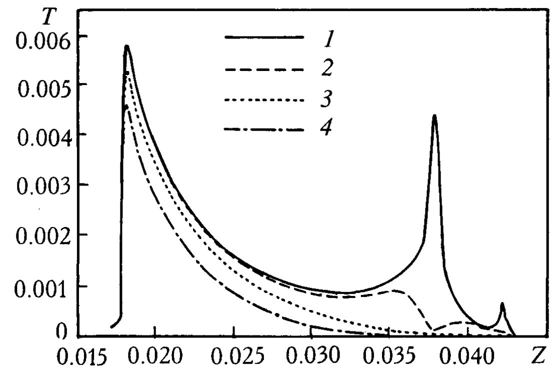

Fig. 3. Temperature distribution at various distances from the optical axis of the eye for a convergence angle of 18^o and an absorption coefficient of 3 cm^{-1}: 1) $r = 0$; 2) $r = 150$; 3) $r = 600$; 4) $r = 1000$ μm. T, ^oC; Z, m.

$$T(r, Z, 0) = T(r_0, Z, t) = 0\,, \quad \frac{\partial T(0, Z, t)}{\partial r} = \frac{\partial T(r, 0, t)}{\partial Z} = \frac{\partial T(r, l, t)}{\partial Z} = 0\,. \tag{2}$$

The radiation absorption in the tissues was determined in accordance with the small-angle approximation of the transfer theory [10], where the heat-release power per unit volume in the *i*th layer was found taking into account the change in the focusing due to both the scattering and the convergence (divergence) of the laser beam incident on the cornea:

$$q_i(r, Z) = \frac{E_{i-1}(k_i + s_i) R_{i-1}^2}{r_s^2} \exp\left[-k_i(Z - Z_i) - \frac{r^2}{r_s^2}\right], \quad r_s^2 = R_{i-1}^2\left(1 + \left(\frac{Z_i}{l_0}\right)^2\right) + \frac{s_i Z_i}{3}(1 - g)\,; \tag{3}$$

$$E_{i-1} = E(Z = Z_{i-1}, r = 0)\,, \quad i = 1, ..., n\,, \tag{4}$$

where l_0 is the distance from the cornea surface to the fundus taken as 24.6 mm; the factor g characterizing the form of the scattering indicatrix is calculated in the Henyey–Greenstein representation [9], and the number of eye tissue layers $n = 6$. The layers in the direction of the laser beam and the coordinates of their boundaries are as follows (numbers between parentheses denote the corresponding elements of the eyes given in Fig. 1):

1) air layer adjacent to the cornea, $Z = 0$ (18);
2) cornea, $Z = 1.77 \cdot 10^{-2}$ m (1);
3) front chamber, $Z = 1.83 \cdot 10^{-2}$ m (16);
4) crystalline lens, $Z = 2.13 \cdot 10^{-2}$ m (2);
5) vitreous body, $Z = 2.53 \cdot 10^{-2}$ m (6);
6) fundus, $Z = 4.23 \cdot 10^{-2}$ m (7, 12, 13).

In the air layer, the heat release is absent: $k_1 = 0$.

The heat-conduction value $\kappa(Z)$ in one of the layers — the fundus tissues — depends on the blood-flow rate. In this problem, the fundus is considered as one of the radiation absorbing layers which actually consists of a thin, strongly absorbing layer of pigment epithelium (Fig. 1, 7) and the choroid lying under it (Fig. 1, 13). According to the data of [11–13], the blood-flow rate in the eye choroid capillaries is much higher that in the other organs. At the same time, for example, in the eye vessels of sugar diabetics a five- to eight-fold decrease in the blood-flow rate is observed [14], which considerably impairs the heat exchange of the choroid. This calculation uses the value of the effective heat conductivity of the choroid characteristic of normal blood circulation: $\kappa = 0.795$ W/(m·deg) [2].

The proposed mathematical model provides the possibility of calculating the temperature fields in all refractive tissues of the eye exposed to the laser radiation of the visible and near IR wavelength ranges with pulse widths from $\sim 10^6$ to 1 sec. The present paper is a development of the mathematical models of the action on the fundus tissues [2, 15] and the cornea of long [3, 4] and short [15] IR radiation pulses that have made it possible to obtain effective

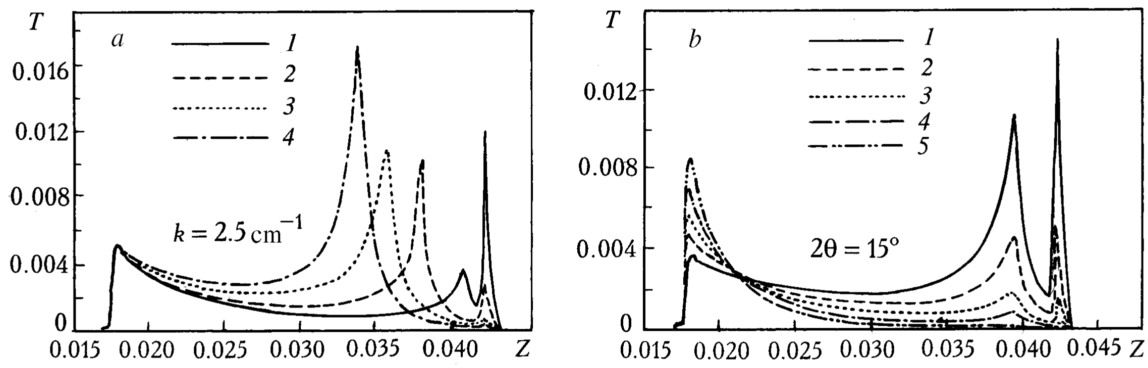

Fig. 4. Axial temperature distribution in the eye tissues under different focusing conditions (a) and at different absorption coefficients (b) of the intraocular refractive media: a) 1) $2\theta = 14$; 2) 16; 3) 18; 4) 20^o; b) 1) $k = 2.0$; 2) 2.5; 3) 3.0; 4) 3.5; 5) 4.0 cm^{-1}.

practical results [17]. Unlike the above-mentioned works, the proposed method and the computation program permit taking into account the beam convergence or divergence angle θ and the radiation scattering coefficient s_i in each layer of refractive tissues of the eye.

In our calculations of the temperature fields in the eye tissues under their heating by laser radiation, we used the values of the absorption coefficients for the tissues of the front section of the eye and for the intraocular media presented in [5, 18]. The values of the thermophysical properties of the eye tissues were taken from [19].

The laser-beam radius on the cornea in the calculations given below is 2 mm. In this case, if the convergence angle of the radiant flux in the vitreous body is $13^o35'$, then the radiation is focused on the fundus. If the convergence angle is larger, the beams are focused in front of the retina. This can cause overheating, for example, in the vitreous body. At certain values of the absorption coefficient the overheating of the transparent intraocular media can be comparable to the overheating of the pigment epithelium of the eye. We estimate the range of parameters at which the temperature values on the cornea surface, at the point of laser focusing, and on the fundus will approximately be of the same order. Since the problem formulated is linear, all calculations have been made for the same radiation power of 0.01 W.

At an absorption coefficient higher than 5–6 cm^{-1} the heating of the fundus tissues is negligibly small due to the strong attenuation of the radiation (by a factor of about 10^7). The limit for the absorption coefficient at which the effects of focusing we are interested in can be observed is no less than $\sim 1\ cm^{-1}$. As to the absorption spectrum of water, which, as shown in Fig. 2, slightly differs from the absorption spectrum of the refractive media, this corresponds to the 1.3–1.4-μm spectral range and (for k of the order of 6 cm^{-1}) to the radiation in the 1.6–1.75-μm range (Fig. 2). To make numerical circulations, we chose the following ranges: the absorption coefficient of the intraocular media $k = (2–5)\ cm^{-1}$ and the convergence angles $2\theta = (13–25)$ deg. The laser-pulse duration in all variants of calculation is taken to be 0.1 sec and the radiation power is 0.01 W. The parameters g and s characterizing the radiation scattering as it propagates, are chosen to be 0.88 and $4.5 \cdot 10^{-3}$ cm^{-1}, respectively.

Figure 3 shows the curve of the temperature distribution along the optical axis of the eye at various radial distances from the axis for a beam convergence angle of 18^o at an absorption coefficient of 3 cm^{-1} corresponding to a wavelength of 1.36 μm. The first temperature maximum is in the cornea layer, the second one — in the vitreous body, and the third one — on the fundus. To the frontal surface of the cornea there corresponds the value of $Z = 1.77 \cdot 10^{-2}$ m. Figure 3 illustrates the possibility of heating localization inside the vitreous body ($Z = 0.038$ m). The sharp decrease in the temperature at $Z > 0.038$ m permits protecting the retinal tissue against photothermal damage. In this case, this is provided by two factors: the decrease in the power (energy) density due to the radiant flux divergence behind the area of focusing and the additional absorption of the radiation of this diverging flux in the medium. The latter fact points to the advantage of near infrared lasers, which up to now has not practically been used.

The dependences of the axial temperature distribution in the eye tissues on the focusing and the absorption coefficient of the refractive media of the eye are shown in Fig. 4. Figure 4a shows the change in the temperatures along the optical axis of the eye at a convergence angle of 15^o and at various absorption coefficients — from 2 to 4

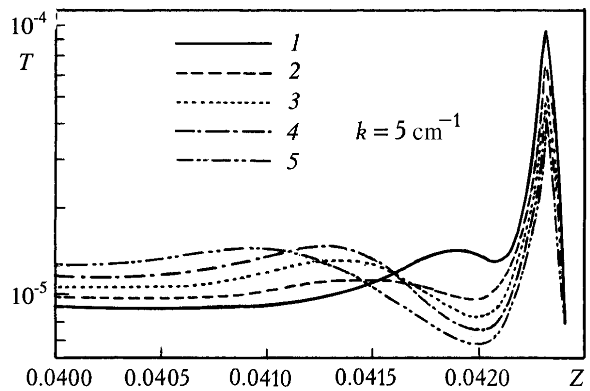

Fig. 5. Temperature field in the vicinity of the fundus upon sharp focusing: 1) $2\theta = 13.4$; 2) 13.5; 3) 13.6; 4) 13,7; 5) 13.8°.

cm^{-1}, and Figure 4b — at an absorption coefficient of 2.5 cm^{-1} and focusing of 14, 16, 18, and 20°. From their comparison one can see well the evolution of the temperature maxima inside the vitreous body of the eye with a change in these parameters. In the general case, to each convergence angle of the beam (in other words, to the spatial location of the pathological object to be irradiated) we can assign the value of the absorption coefficient (radiation wavelength) at which a given level of fundus protection (the irradiation level corresponding to the maximum permissible level (MPL)) at a simultaneous local heating of the medium is provided. In so doing, the variations of the absorption coefficient are small and lie within the range of ~2–5 cm^{-1}.

All calculations of the temperature fields have been performed for the case of clean intraocular media, i.e., we assumed the absence from the vitreous body of hemorrhages or vascular neoplasms. In the presence of such a pathology, the absorption coefficients at the focusing point sharply increase to provide an additional possibility of thermal coagulation. This offers a fundamental possibility of effective laser treatment of neoplasms characteristics of diabetic retinopathy, hemorrhages in the vitreous body, and other forms of pathology with a simultaneous increase in the level of protection of the sensor layers of the retina.

The 1.6–1.75-μm range, to which there correspond absorption coefficients of the order of 5–6 cm^{-1}, is promising for laser surgical operations in the front section of the eye (Fig. 1, 1–5, 15, 16), since throughout the cornea-retina portion the decrease in the power (energy) density exceeds 10^5. In this case, the following fact deserves attention. Despite such a high attenuation, the focusing of the luminous flux immediately on the fundus can cause a considerable local heating of the pigment layers of the retina (Fig. 5), exceeding several times the cornea heating. Incidental irradiation of the eye by a collimated luminous flux in the 1.6–1.75-μm spectral range considerably increases the probability of thermal damage of the retina. This fact is ignored by both domestic [20] and foreign normative documents defining the conditions for safe use of lasers. Laser radiation with a wavelength larger than 1.4 μm is traditionally considered as "safe" to the retina, and the MPLs of irradiation of the eyes by directional luminous fluxes are determined in this case from the conditions providing cornea safety. As follows from the results of the calculations presented in Fig. 5, in the case under consideration such an approach is unacceptable or, at least, requires serious additional substantiation.

Thus, the proposed approach to the calculation of the dynamics of the temperature field induced by a powerful optical radiation in the internal structural elements of the eye has made it possible to reveal a number of features of the action on the eye of near infrared laser radiation that are of interest for both clinical ophthalmology and estimation of the danger of such a radiation.

This work was supported by the Belarusian Republic Basic Research Foundation (project F99P-056).

NOTATION

Z and r, axial and radial coordinates; λ, laser radiation wavelength; $\kappa(Z)$, heat conductivity coefficient; $T(r, Z, t)$, excess of the tissue temperature over the physiological norm (37°C); $\rho(Z)$ and $C_p(Z)$, density and heat capacity, respectively; $Q(t)$, time dependence of absorbed radiation intensity; q_i, heat-release power per unit volume in the ith layer; n, number of tissue layers; k and s, absorption and scattering coefficients; E, irradiation of the front boundary of the ith layer on the laser-beam axis; R_0 ($i = 1$), light-beam radius on the intensity level exp (–1) from maximum; r_s, light-beam radius with allowance for the scattering and divergence; 2θ, convergence angle of the light beam focused on the retina by the optical system of the eye; g, factor characterizing the form of the scattering indicatrix.

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