Optical characteristics of cartilage at a wavelength of 1560 nm and their dynamic behavior under laser heating conditions

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Abstract. A double-integrating-sphere system was used to measure the diffuse transmittance, diffuse reflectance, and collimated transmittance of cartilage and polyacrylamide hydrogel samples as a function of temperature under 1560-nm laser heating conditions. The dynamic behavior of the absorption and scattering coefficients and scattering anisotropy of the biomaterials was calculated by the inverse Monte Carlo method. The absorption coefficient of the cartilage and hydrogel samples proved to be linear in temperature. Raising the temperature of the cartilage samples to 80°C caused their absorption coefficient to decrease by some 25%. The temperature-induced change of the absorption spectrum of the interstitial water was found to be responsible for the clarification of the cartilage tissue observed to occur under 1560-nm laser heating conditions. The temperature field produced in the tissue by the laser energy deposited therein was calculated using a bioheat transfer equation with temperature-dependent parameters. The calculation results demonstrated that the temperature-induced changes of the optical parameters of biological tissues should be taken into account to make their 1560-nm laser treatment effective and safe. [DOI: 10.1117/1.3484749]

Keywords: optical properties; laser heating; dynamic behavior; cartilage; polyacrylamide gel; IR spectrum of water; temperature dependence.

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1 Introduction

A number of treatment procedures have recently been developed on the basis of a moderate thermal heating of biological tissues with laser radiation.[1] The principal merit of laser radiation is the possibility it offers to convey energy to a local region in the biotissue being treated as required by the given scenario. To provide for the necessary therapeutic effect and safe conduct of the medical procedures involved, one should strictly control the temperature field induced in the tissue by the laser. This can only be done if one knows the optical and thermophysical characteristics of the tissue and their variations during the course of laser treatment. The temperature field is determined by both the parameters of the laser radiation proper, namely, its wavelength, the energy distribution in the laser beam, and exposure time, and the optical characteristics of the tissue at the laser wavelength. The set of the three optical parameters—the absorption coefficient μ_a, scattering coefficient μ_s, and anisotropy factor g at a given scattering phase function—wholly determines the distribution of light in the bulk of the tissue, and hence, the heat-source function associated with light absorption. It characterizes the rate of temperature rise in the laser-exposed zone and controls the spatial and temporal configurations of the temperature field. These optical parameters are usually determined by indirect methods based on room-temperature measurements of diffuse reflection and transmission of low-intensity radiation,[2–7] taken with integrating spheres. The results obtained are subsequently used to describe the propagation of high-intensity laser radiation in the biological tissue of interest on the assumption that the radiation and the tissue both remain unchanged during the course of laser heating.

However, the thermal action of laser radiation on biological tissues gives rise therein to a great many physicochemical processes, at both the microscopic and macroscopic levels.[8] As a result, changes in the structure and composition of the tissues take place that can lead to substantial alterations in their optical parameters at the laser wavelength. For example, the anisotropy factor and scattering coefficient usually change as a result of either conformation transitions of the macromolecules forming the biological tissue or its thermal coagulation.[9–15] The absorption coefficient depends on the concentration of the chromophore in the laser-exposed zone. Where the chromophore is water, the reduction of the absorption coefficient is sometimes attributed to the thermal diffusion of water from this zone.[16–18] In a number of cases, the

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absorption dynamics of biological tissues during laser action thereon is governed by the temperature dependence of the absorption spectrum of the intratissue water. Numerous facts are available today in the literature that evidence variations of the absorption spectrum of liquid water under the effect of various lasers emitting radiation in the wavelength range from the UV to the far-IR region. To illustrate, Ediger et al.[19] demonstrated that when ablating hydrated collagen by the ArF laser radiation ($\lambda=193$ nm), its absorption of radiation increased materially even during a short pulse (~ 25 ns). Later on it was found[20] that this effect was due to a change in the absorption spectrum of water in the UV region upon heating. Jansen et al.[21] investigated the transmission dynamics of a water layer for the Ho:YAG ($\lambda=2.12$ μm), Ho:YSGG ($\lambda=2.09$ μm), and Tm:YAG ($\lambda=2.01$ μm) laser radiation at various temperatures. They showed that when water was heated to 70°C, the absorption of radiation by it lowered perceptibly as a result of the shift of its $2\nu_2$ absorption band (1.94 μm) toward the short-wavelength region. The clarification of pure water was also observed to occur under the effect of high-power Er:YAG (2.94 μm) and Er:YSGG (2.79 μm) laser pulses,[22–24] when its temperature reached a few thousand degrees centigrade. The experimental results presented by Shori et al.[22] and Walsh and Cummings,[24] together with their respective coworkers, confirmed the shift of the ν_1 (3.05 μm) and ν_3 (2.87 μm) absorption bands of liquid water toward shorter wavelengths as a result of the weakening of the hydrogen bond between the water molecules on heating. They suggested models for crater formation in the laser ablation of biological tissues that allowed for this absorption band dynamics. A similar dynamics was also observed by Sobol et al.[25] who studied the radiometric response of cartilage tissues and cornea to the action of a tunable free electron laser.

In recent times, there has been tremendous progress in the development of medical fiber and solid state erbium glass lasers emitting radiation in the wavelength range 1.54–1.56 μm. They are being extensively used in medical practice for correction of cartilage tissues,[26,27] skin rejuvenation by multiple local laser heating actions (fractional photothermolysis),[28,29] and spine disk regeneration.[30] This has stimulated investigations into the mechanisms governing the interaction between the above-mentioned laser radiation and biological tissues. Bagratashvili et al.[31] demonstrated earlier that cartilage tissues undergo reversible clarification upon laser heating at a wavelength of 1.56 μm. Several mechanisms were suggested that could explain such a clarification. First of all, this can be due to the effect of the thermal diffusion of water from the laser-exposed zone and deformation of the tissue absorption spectrum upon heating (absorption band shift and reduction of the integral absorption intensity as a result of the deaggregation of water clusters upon heating). All the mechanisms suggested were adequately substantiated,[32] but their individual contributions to the tissue clarification effect observed were not determined. It is important to note that the dynamics of the transmission of laser radiation through the cartilage tissue studied was measured in the above work[31] with a receiver of 10 mm in diameter located immediately behind the tissue sample, which provided for a sufficiently large angular aperture for the collection of the transmitted light and, hence, for the adequate reliability of the energy measurements. However, the experimental scheme used was incapable of measuring the optical parameters of the cartilage tissue and their dynamics during the course of laser treatment. At the same time, such measurements, on the one hand, are of practical importance in selecting optimal laser treatment conditions, and on the other hand, they can help elucidate the mechanism behind the clarification of cartilage tissues at a laser wavelength of 1.56 μm.

In essence, the present work was a continuation of the work by Bagratashvili et al.[31] Its objective it was to measure the optical parameters of cartilage tissue and polyacrylamide hydrogel samples at a wavelength of 1.56 μm and study their dynamics in the course of laser heating. Subject to consideration here is the problem of safety in medical laser procedures associated with the tissue clarification effects. The results obtained can be extended to other biological materials as well.

2 Materials and Methods

The samples studied were prepared from calf nasal septal hyaline cartilage. The tissue was freed from perichondrium and mucous coat. The samples were prepared the day the experiment was conducted and stored in saline at a temperature of 4°C. Such a sample storage method minimizes the tissue decomposition as a result of natural decay processes.[33]

Polyacrylamide (PAA) hydrogel was selected to serve as a model system. This synthetic material is being frequently used as an equivalent of biological tissues.[34] Its merits are good elasticity and thermal stability at temperatures up to 90°C. The PAA hydrogel is convenient to use because its samples can be synthesized to differ in water content and degree of cross-linking. The PAA hydrogels were synthesized through radical copolymerization of acrylamide (Lancaster Synthesis, Windham, New Hampshire) and N,N′-methylenebisacrylamide (Amresco, Solon, Ohio) in the presence of catalytic amounts of ammonium persulfate (Amresco) and N,N,N′,N′-tetramethylenediamine (Lancaster Synthesis).[35] Using this method, we synthesized a PAA hydrogel 1:19 in cross-linking degree and 70% by mass in water content that possessed optical and thermophysical characteristics similar to those of the cartilage.[34]

To find the temperature dependence of the optical parameters, we used a measuring system including a radiation source, two integrating spheres 8 cm diam, and a sample temperature measuring unit (Fig. 1). The inside surface of the spheres was coated with fine-dispersed barium sulfate (Kodak, Los Angeles, California), ensuring high reflectance in the optical and near-IR regions. The radiation intensity on the inside surface of each sphere was measured with InGaAs/InP photodiode (Telam, Moscow, Russia) operating in the spectral range from 1100 to 1620 nm. The diameters of detector ports were equal to 5 mm. Investigations into the integrating capacity of the spheres showed that variations in the direction of the 1560-nm radiation entering the sphere had practically no effect on the sensor signal being measured. Each sphere had two ports opposite to each other. The diameters of both ports were equal to 1 cm. Both spheres were mounted on movable supports so that their ports could be aligned. One more photodiode sensor was installed on the same axis at a distance of ~ 120 cm from the exit port of the second sphere. The data from all the sensors were collected with a Model USB-6009

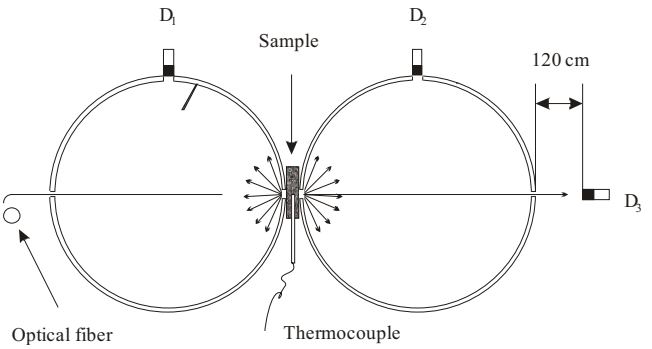

Fig. 1 Schematic diagram of the double-integrating sphere setup for measuring diffuse transmittance, diffuse reflectance, and collimated transmittance.

multichannel analog-to-digital converter (National Instruments, Austin, Texas). The dark noise level amounted to 0.002 V at a characteristic signal level of 1–1.5 V. On the basis of these data and using calibration coefficients, we calculated the diffuse reflection (R_d) and diffuse transmission (T_d) intensities, and also the intensity of light passing through the sample without deflection from the original direction (T_c). The measurement accuracy of these quantities came to 0.2, 0.3, and 6×10^{-5} mW, respectively.

The radiation source used was a fiber erbium glass laser (IRE-Polus, Moscow, Russia) 1.56 μm in radiation wavelength and up to 5 W in output power. The transmission of radiation was by means of a quartz optical fiber 0.6 mm diam. We measured the spatial light intensity distribution at the exit from the fiber at different distances from its end face by scanning the light beam with a sensor whose sensing element was 25 μm across. These measurements showed that the intensity distribution in the near and far fields could be well described by a Gauss formula. The exit face of the optical fiber was fixed inside the first integrating sphere at a distance of 7 mm from the sample surface. The intensity distribution in the sample plane was described by

$$I(r) = I_0 \exp\left(-\frac{r^2}{W_L^2}\right), \quad (1)$$

where $W_L = 0.8$ mm is the radius of the laser beam and r is the distance from its center.

The cartilage tissue or PAA hydrogel sample of specified thickness ($\sim$1.5 mm) was placed between two 1-mm-thick glass plates. The sample temperature was measured with a K-type thermocouple (Omega Engineering, Stamford, Connecticut) 125 μm in thickness that was inserted into the sample in order to locate its junction at the center of the radiation-exposed zone. Thanks to its small thickness, the thermocouple had practically no effect on optical measurements. Furthermore, under laser heating conditions, it was practically always in thermal equilibrium with the surroundings. Our detailed investigation of temperature dynamics at the beginning and end of laser pulse showed that the temperature jump caused by direct heating of thermocouple by laser light at 1.56 μm, does not exceed 1.5 °C. We consider this value to be rather small and acceptable for our experiments. The data from the thermocouple were collected with a model no. cDAQ-9172, 9211 analog-to-digital converter (National Instruments). The random error of temperature measurement was 0.1 °C.

Thus, the experimental setup described allowed us to synchronously measure, in real time, the diffuse reflectance R_d, diffuse transmittance T_d, collimated transmittance T_c, and sample temperature in the radiation-exposed zone.

2.1 *Calculation of Optical Parameters*

The diffuse reflectance R_d, diffuse transmittance T_d, and collimated transmittance T_c were calculated on the basis of a Monte Carlo algorithm similar to that used by Wang et al.[36] and Pope and Wang.[37] The photon flux reaching the sensitive element of the third sensor was calculated with due consideration for the angular distribution of the incident beam and the intensity distribution on the sample surface. The Henyey–Greenstein function generally accepted in biomedical practice was used to serve as a scattering phase function; 3×10^5 iterations of the Monte Carlo program were needed for calculation of the R_d, T_d, and T_c parameters.

The determination of the unknown parameters μ_s, μ_a, and g reduced to the search for their values at which the sum of the squared deviations of the theoretical quantities R_d^{theor}, T_d^{theor}, and T_c^{theor} from their experimental counterparts R_d^{exp}, T_d^{exp}, and T_c^{exp} was minimal. This problem was solved by the method of linear regression using the modified Levenberg–Marquardt algorithm.[38] The quantities R_d^{exp}, T_d^{exp}, and T_c^{exp} for each sample were measured five times, which made it possible to calculate five statistically independent groups of the parameters μ_s, μ_a, and g and the corresponding standard deviations $\sigma\mu_s$, $\sigma\mu_a$, and σ_g that came to 0.4, 0.1, and 0.01 cm^{-1}, respectively. In these calculations, the values of μ_s, μ_a, and g were taken to be the same throughout the bulk of the sample.

3 Experimental Results

Using the above-described measuring system, we found the temperature dependences of R_d, T_d, and T_c for our cartilage tissue and PAA hydrogel samples under laser heating conditions. Characteristic results for the cartilage tissue are presented in Fig. 2. As the laser heated the cartilage samples to 80 °C, their R_d, T_d, and T_c rose monotonically by approximately 5, 65, and 60%, respectively. The R_d, T_d, and T_c values measured for the PAA hydrogel samples proved close to those for their cartilage tissue counterparts, as were their temperature dependences. The R_d, T_d, and T_c values measured following the spontaneous cooling of the samples to room temperature and their repeated laser heating under the same conditions practically coincided with the values measured previously for both the cartilage tissue and PAA hydrogel samples. It should be noted that the slope of the R_d temperature-dependence curve for the cartilage tissue somewhat decreased in the temperature range from 45 to 60 °C, this being true of all the samples and during the course of the repeated laser heating cycle as well. At the same time, no such deviations from linearity were observed in the PAA hydrogel samples. No R_d measurements were taken for water. Doubling the laser radiation intensity increased the heating rate of the

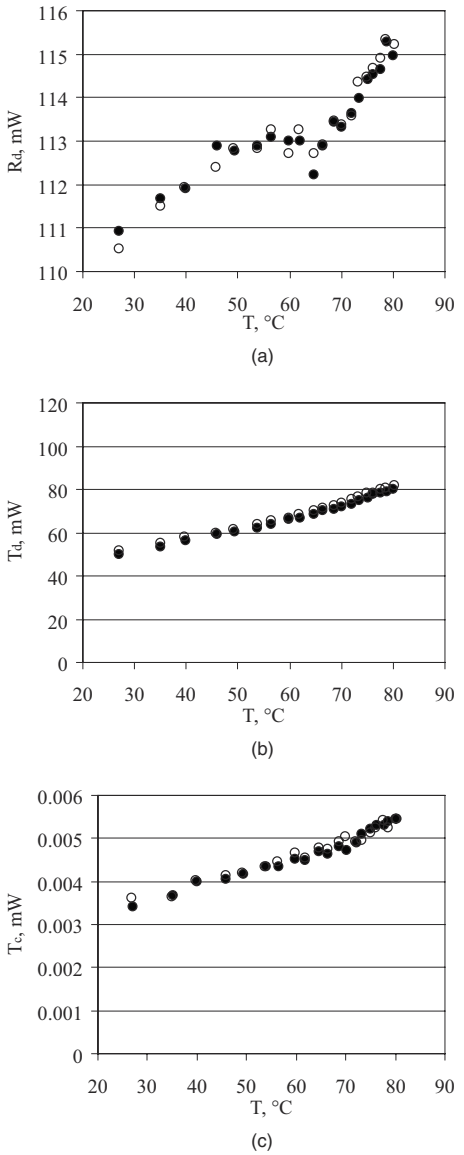

Fig. 2 Temperature dependences of (a) R_d, (b) T_d, and (c) T_c for the cartilage tissue (● — first laser heating cycle; ○ — repeated laser heating cycle).

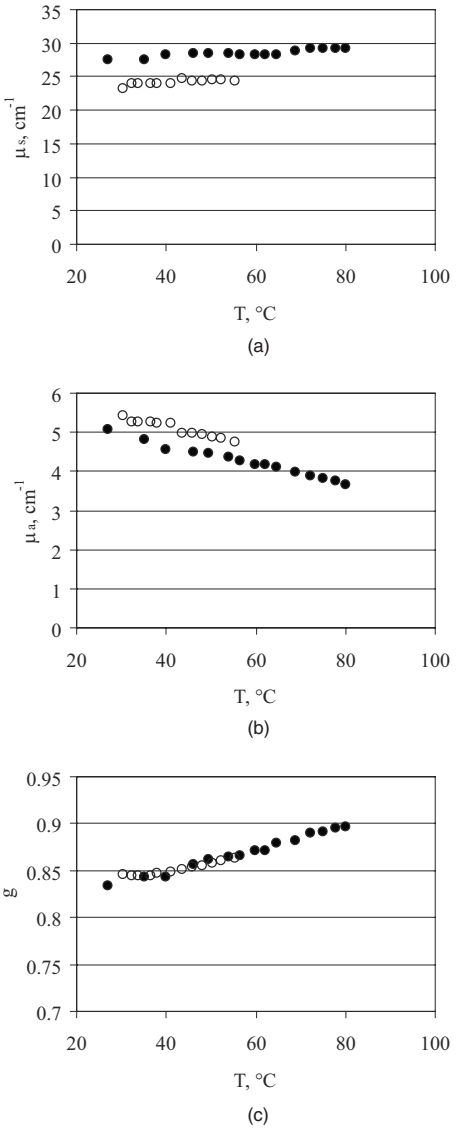

Fig. 3 Temperature dependences of the coefficients (a) μ_s, (b) μ_a, and (c) g for the cartilage (●) and PAA hydrogel (○).

samples by two times, but had practically no effect on the character of the temperature dependence of R_d, T_d, and T_c.

The experimental results obtained for R_d, T_d, and T_c allowed us to calculate the corresponding optical parameters: the scattering coefficient μ_s, absorption coefficient μ_a, and anisotropy factor g of the cartilage and PAA hydrogel. The temperature dependences of these parameters are presented in Fig. 3. It can be seen that the coefficient μ_s for both materials varies within 5%, its increase being certainly monotonic. The anisotropy factor also increases monotonically and practically linearly from 0.83 to 0.9 as the cartilage sample is heated from 20 to 80°C. The absorption coefficient of the cartilage at room temperature is ~5 cm^{-1}, and during the course of laser heating, it decreases linearly, the slope of its temperature curve amounting to some 0.025 $cm^{-1}/°C$. Thus, it decreases to 3.6 cm^{-1} upon laser heating to 80°C, which comes to some 75% of its original value. Approximately similar values of the coefficient μ_a and the same slope of its temperature curve are also observed for the PAA hydrogel. For comparison, we measured the temperature dependence of the absorption coefficient of liquid water during the course of laser heating at a wavelength of 1.56 μm by a similar scheme using a liquid cell with a built-in thermocouple. In that case, the exit port of the second integrating sphere was shut with a standard diffuse reflector. The results obtained are presented in Fig. 4. The absorption coefficient of water measured at room temperature is ~12 cm^{-1}, which agrees with the available literature data.[39] One can see that the absorption coefficient of water also suffers a linear drop; its relative change agrees well with similar variations for the cartilage and PAA samples under the same heating conditions, with due regard for their water content, and amounts to some 0.005 $°C^{-1}$.

4 Discussions

Our measurements showed that the laser heating of the cartilage tissue and PAA hydrogel from 25 to 80°C caused their

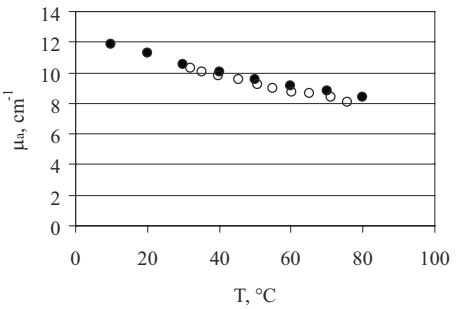

Fig. 4 Temperature dependence of the coefficient μ_a of liquid water (● — literature data;[40] ○ — experimental data).

optical parameters to undergo reversible changes. The reversible character of these changes is evidence that no irreversible processes, such as coagulation[10,13] or dehydration,[17] took place in the samples under our laser heating conditions. We consider the following physicochemical processes as possible causes of the effects observed:

1. the formation of a thermal lens in the sample
2. mass transfer of water contained in the biological tissue or hydrogel as a result of the laser-induced temperature gradient therein
3. reversible conformation transformations of the cartilage and hydrogel macromolecules
4. alteration of the absorption spectrum (shape and integral intensity) of water upon heating

Note that these processes can occur simultaneously, each being capable of contributing to the alteration of the optical properties of biological tissues.

The laser heating of a biological tissue sample induces therein an inhomogeneous temperature field that can generally give rise to two phenomena: formation of a thermal lens and mass transfer of water contained in the biomaterial from the center of the laser-exposed zone to the periphery. On the one hand, the laser-induced temperature gradient produces a refractive index gradient along the laser beam path in the biomaterial, which can affect the readings of the sensors, especially those of the sensor used to measure T_c.[13,16] On the other hand, the presence of a temperature gradient causes some redistribution of water—the principal 1.56-μm chromophore of biomaterials—in the laser-exposed zone, which can also affect the readings of the sensors used to measure T_d and T_c. But in our case, these effects are apparently unimportant. Both of these effects depend on the magnitude of the temperature gradient in the laser-exposed zone and ought, as a first approximation, to be linear in radiation power. Doubling the laser power in our experiments caused no perceptible changes in T_d and T_c, provided that the sample temperature remained the same. This means that thermal lens formation and water redistribution in the laser-exposed zone have no significant effect on the R_d, T_d, and T_c values being measured, hence on the optical parameters determined on their basis.

The similarity between the temperature dependences of the coefficients μ_s and g for the cartilage and polyacrylamide hydrogel bears witness to the fact that their changes are primarily caused by the alteration of the state of water contained in the samples. However, the slight difference between the $\mu_s(T)$ curves for the cartilage and PAA hydrogel in the temperature range 45–60°C points to individual specificities of these systems. It is well known that hydrogels as well as biological tissues contain kinetically and thermodynamically unequal water molecules.[40] The nuclear magnetic resonance spectra of such systems show two lines[41,42] instead of a single one, as is the case with liquid water. This fact points to the existence of water specifically linked to the polymer and also water not linked to the polymer ("free water"). There is also information that water changes from one state to the other upon heating. Therefore, the changes observed in μ_s and g are possibly due to a gradual change of the "linked" water to the "free" state under the effect of laser heating, as well as conformation changes of macromolecules consequent on changes in temperature and water state.

Worthy of note is the hypothesis that the laser heating of biological tissues and PAA hydrogels causes their dissolved gas to liberate in the form of bubbles, which increases light scattering. Indeed, the solubility of gases in water decreases with rising temperature. If their concentration is close to saturation, some dissolved molecules must change to the gaseous state upon heating. Obviously this process can take place without any temperature-controlled threshold, and in a reversible manner and rather rapidly. But to confirm this hypothesis requires additional investigation.

The substantial reduction of the absorption coefficient μ_a of the cartilage and PAA hydrogel with rising temperature is apparently due to the change of the absorption coefficient of liquid water. That this is so, is supported by both our measurements of the total transmission of liquid water upon laser heating (Fig. 4) and the well-known investigations into the deformation of the 1470-nm absorption band $(\nu_1+\nu_3)$ of liquid water upon ordinary heating under equilibrium conditions.[43] Indeed, the relative changes of the dependence $\mu_a(T)$ for the cartilage, PAA hydrogel, and water practically coincided at equal temperatures, which points to a common nature of the phenomenon. As follows from the work by Segtnan et al.,[43] as temperature grows higher, the absorption band maximum of liquid water gradually shifts toward the short-wavelength region. These authors believe that such a deformation of the absorption band (1470 nm; $\nu_1+\nu_3$) of water upon ordinary heating under equilibrium conditions is due to the equilibrium change of one type of water clusters formed by several water molecules to another type of cluster consisting of a fewer number of molecules. This mechanism is probably also applicable to the description of the temperature dependence of the coefficient μ_a of any biomaterials, including cartilage tissues and PAA hydrogels. Our analysis of the above work[43] showed that the temperature dependence of μ_a of water at a wavelength of $\lambda=1.56$ μm proved practically linear, the slope of its curve being in good agreement with our measurements (Fig. 4).

When normalized to water concentration, the values of the coefficient μ_a measured for the PAA hydrogel and cartilage samples at room temperature turned out to be somewhat lower than the expected 7–8 cm^{-1}, namely, 5–5.5 cm^{-1}. This difference can be due to various factors.

First, account should be taken of the fact that the optical parameters determined are the result of solution of a complex inverse problem involving a great many parameters and a number of assumptions. For example, the use of the Henyey–

Greenstein scattering phase function in computing these parameters is, in principle, unfounded. This function is of a phenomenological character but offers a number of advantages and conveniences. In many cases, it yields reasonable results, provided that the experimental data obtained are used to calculate other quantities, for example, the radiation power density distribution inside the scattering medium.

Second, some systematic error in determining the optical parameters can arise in measuring and modeling T_c. In our model, we tried to take account of the actual geometrical factors of the experimental scheme, the angular and spatial light distribution in the laser beam incident on the sample. However, our model assumed a geometrical character of light propagation and failed to allow for possible coherence effects affecting the wavefront in the near and far fields.

Third, the fact cannot be ruled out that the spectral properties of pure water can actually differ perceptibly from those of water contained in biological tissues and PAA hydrogels. Therefore, evaluating the value of the coefficient μ_a as the product of the value of this coefficient for liquid water by the water concentration corresponding to the biomaterial in hand is conditional enough. Calculations by the Monte Carlo method of the total diffuse transmission T_d as a function of the sample thickness h (measured in centimeters) at measured optical parameters and laser beam parameters corresponding to our experiment yielded the exponential dependence $T_d = K \exp(-\mu_{eff} h)$, where $\mu_{eff} = 11.6\ \mathrm{cm}^{-1}$. The diffusion length $\mu_d = \sqrt{3\mu_a[\mu_a + (1-g)\mu_s]}$ of photons wandering in the cartilage at room temperature amounted to $11.8\ \mathrm{cm}^{-1}$ at the parameters $\mu_a = 5\ \mathrm{cm}^{-1}$, $\mu_s = 28\ \mathrm{cm}^{-1}$, and $g = 0.85$, measured by us at room temperature. The coefficients μ_a and μ_{eff} agree well with the effective absorption coefficient of the cartilage tissue, $\mu_{eff} = 10.8 \pm 0.5\ \mathrm{cm}^{-1}$, measured independently by us[34] by the method of remote laser IR radiometry. These facts indirectly bear witness to the reliability of the values of the optical parameters obtained.

When calculating the optical parameters, we assumed that the values of μ_s, μ_a, and g were the same throughout the bulk of the sample. But in actual fact, with the temperature field inside the sample being inhomogeneous and the quantities μ_s, μ_a, and g, temperature dependent, one cannot strictly state that they remain the same for all points of the sample upon laser heating. In the ideal case, to determine μ_s, μ_a, and g, one ought to solve a complex inverse problem, with due consideration given to the mutual influence of the light and temperature fields under varying optical parameter conditions. The problem is solvable, given specified *a priori* temperature dependences of μ_s, μ_a, and g, but requires substantial complication of the light and heat propagation models available at the given moment. In this work, we did not attempt such a task, restricting ourselves to the approximation presented above. Apparently such an approach was justified, for the temperature dependences of μ_s, μ_a, and g proved rather gradual and the temperature gradients proper in the region wherethrough the bulk of photons passed were not very great.

The variation of the optical parameters observed during the course of laser heating to 80 °C at a wavelength of $\lambda = 1.56\ \mu\mathrm{m}$ leads to a 30% increase in the depth of penetration of laser radiation, which is important in selecting irradiation regimes as regards their efficiency and safety. To evaluate the effect of the dynamics of μ_s, μ_a, and g on the temperature field in biological tissues, use was made of the model presented in our work.[34] The calculation of the laser-induced three-dimensional temperature field was based on the solution of the classical heat conduction equation expressed in cylindrical coordinates,

$$\frac{\partial T}{\partial t} = \chi \frac{1}{r}\frac{\partial}{\partial r}\left(r\frac{\partial T}{\partial r}\right) + \chi \frac{\partial^2 T}{\partial z^2} + f(r,z,t), \quad 0 \leq r \leq R, \quad 0 \leq z \leq Z, \tag{2}$$

where $T(r,z,t)$ is the temperature at the point with the coordinates (r,z) at the instant t, $f(r,z,t)$ is the laser heating rate (in Kelvin per second) of the medium, and $\chi = 0.13$ is its thermal diffusivity (in millimeters squared per second). The radiation intensity inside the sample was assumed to be described by the Beer law; hence, the function $f(r,z,t)$ for light beams with a Gaussian energy distribution was found to be

$$f(r,z,t) = (1 - R_d)\frac{\mu_{eff} P}{\rho C_p \pi W_L^2} \exp\left[-\left(\frac{r}{W_L}\right)^2\right] \times \exp(-\mu_{eff} z) \cdot \theta(t), \quad \theta(t) = \begin{cases} 1, & 0 < t \leq t_{imp} \\ 0, & t > t_{imp} \end{cases}, \tag{3}$$

where μ_{eff} is the effective laser radiation attenuation index, $\rho C_p = 3.5$ is the specific heat capacity (in Joules per cubic centimeters per degree kelvin), P is the laser power (in watts), $W_L = 0.1$ is the laser beam radius (in millimeters), and R_d is the dimensionless diffuse reflectance measured with an integrating sphere and t_{imp}, is the time that the laser source is turned on. The spatial distribution of the laser power density in an absorbing optically inhomogeneous medium depends on μ_s, μ_a, and g and, strictly speaking, is not amenable to the Beer law. Nevertheless, the use of an exponential law with some generalized index μ_{eff} to describe the distribution of heat sources in an optically inhomogeneous absorbing medium seemed quite justified. We also assumed that $\mu_{eff}(T) = \mu_d(T) = \sqrt{3\mu_a(T)\{\mu_a(T) + [1 - g(T)]\mu_s(T)\}}$, where the dependences $\mu_a(T)$, $\mu_s(T)$, and $g(T)$ corresponded to their measured counterparts. It turned out that the dependence $\mu_{eff}(T)$ could be well described by the straight line $\mu_{eff}(T) = -0.065T + 13.8$, where μ_{eff} is measured in inverse centimeters and temperature (in degrees Celsius).

The temperature of the sample at the initial instant of time was taken to be homogeneously distributed throughout its bulk and equal to the ambient temperature T_0. Boundary conditions of third kind were specified at the boundary of the sample. The problem formulation provided for the independence of thermal diffusivity and specific heat capacity from temperature, coordinates, and time.

Within the scope of these notions, the solution of the thermal problem by the finite difference method showed a substantial difference between the temperature fields produced upon laser heating in cartilage tissues with the coefficient μ_{eff} remaining steadily at $11.8\ \mathrm{cm}^{-1}$ and with that varying linearly with temperature. The temperature was computed at each finite difference step of the thermal model, and then the optical properties adjusted according to the new temperatures to de-

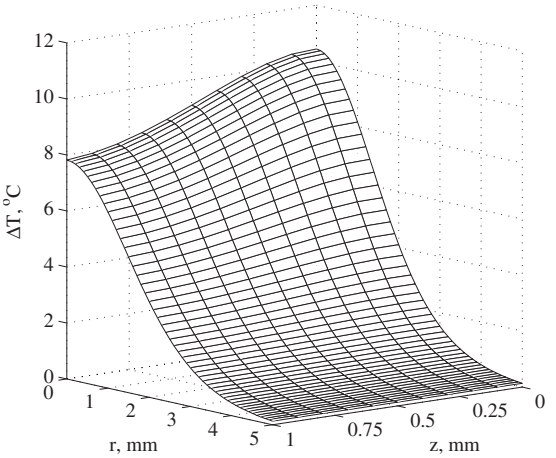

Fig. 5 Difference between temperature fields calculated with and without allowance for the optical parameter dynamics.

termine a new light distribution based on Eq. (3) with the process repeated for each time subsequent step. Figure 5 presents the distribution of this temperature difference as a function of sample depth and distance from the laser beam axis, calculated at a radiation power of 1 W, laser spot radius of W_L=0.8 mm, and heating time of 6 s, which approximately corresponds to conditions typical of laser nasal septum reshaping procedures. As one can see, the temperature difference on the front surface of the sample reaches some 12°C, this on the opposite surface amounting to 8°C. Such a temperature difference is important and should be taken into account when selecting laser treatment conditions for biological tissues.

5 Conclusions

Short-term 1.56-μm laser heating of cartilage tissue to 80°C causes its optical parameters to undergo reversible changes. The values of the coefficients μ_a, μ_s, and g at room temperature come to 5.1 ± 0.1, 27.5 ± 0.4, and 0.83 ± 0.01 cm^{-1}, respectively. As temperature rises to 80°C, the coefficients μ_s and g increase monotonically by 6–8%, while μ_a decreases by some 25%. The dependence $\mu_a(T)$ is almost linear, with the slope of its curve amounting to 0.025 cm^{-1}/°C.

The reduction of the coefficient μ_a observed to occur with rising temperature is due to the laser-induced change of the absorption spectrum of water contained in the biomaterials. The temperature dependences of the coefficient μ_a of the cartilage and PAA hydrogel, normalized to their water content, are characterized by one and the same slope, similar to that of liquid water, which is evidence of the same nature of the phenomenon in these materials.

Modeling the temperature field dynamics with and without allowance for the temperature dependences of the coefficients μ_a, μ_s, and g showed that the maximum difference between the respective temperature fields produced in the cartilage tissue upon its laser heating to 80°C in conditions typical of laser reshaping procedures reached 12°C. Allowing for the tissue clarification process increased the penetration depth of laser radiation. From these considerations, it is recommended that one should take account of the temperature dependence of the optical parameters of the cartilage tissue when modeling the temperature fields induced therein by 1.56-μm laser radiation and selecting laser treatment conditions for biological tissues in practice.

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