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# Equivalent of a cartilage tissue for simulations of laser-induced temperature fields

A.V. Kondyurin, A.P. Sviridov

***Abstract*. The thermal and optical properties of polyacrylamide hydrogels and cartilages are studied by the method of IR laser radiometry. The thermal diffusivity, heat capacity, and the effective absorption coefficient at a wavelength of 1.56 μm measured for polyacrylamide gel with 70 % water content and the degree of cross-linking 1 : 9 and for the nasal septum cartilage proved to be close. This allows the use of polyacrylamide hydrogels as equivalents of cartilages in simulations of laser-induced temperature fields.**

***Keywords***: *laser radiometry, temperature field, thermal properties, phantoms of biological tissues, polyacrylamide gel, cartilage.*

## 1. Introduction

Artificial materials imitating individual physicochemical properties of biological tissues are more and more often used to simulate a response caused by various physical actions. These materials are stable, convenient in applications, allow one to control easily doses and action regimes, and considerably simplify the development and tests of the diagnostic medical equipment. Depending on the type of action, it is necessary to reproduce various physical properties of biological tissues. At present many phantoms are developed and used to imitate the optical [1 – 3], acoustic [2 – 4], electric [5 – 7], and thermal [6 – 8] properties of biological tissues.

Of special interest are equivalents of biological tissues used to simulate laser-induced heating of tissues. In particular, a moderate heating of biological tissues by laser radiation up to 70 – 80 °C is used for cartilage shaping [9 – 11], stimulation of regenerative processes, and hyperthermia of malignant tumours. To achieve the therapeutic effect, the temperature field should be produced in strictly specified temporal and spatial limits. The spatiotemporal temperature distribution in a biological tissue is determined by the parameters of laser radiation and the absorption and scattering coefficients, specific heat and thermal diffusivity of the tissue. It is obvious that materials with optical and thermal properties similar to those of biological tissues can be used as their equivalents for simulating laser-induced heating. It is important that possible thermal and chemical processes would not distort considerably the temperature field and the phase stability should be preserved.

The aim of our study is to obtain the equivalent of a cartilage tissue for simulating temperature fields induced by near-IR laser radiation. The main component of a cartilage tissue absorbing radiation in this spectral range is water (70 % – 80 %). For this reason various hydrogels are most convenient for fabricating the required equivalent. Among them are agar [12], agarose [13], gelatine [14], and polyacrylamide (PAA) [5 – 7] hydrogels. However, most of these hydrogels melt already upon slight heating and only the PAA hydrogel preserves its elasticity, which makes it possible to synthesise samples of various shapes and to introduce probes and thermocouples into PAA. By synthesising PAA gels with different monomer concentrations and different degrees of cross-linking, we can obtain a material with thermal and optical properties similar to those of a cartilage tissue.

In this paper, we studied by the IR radiometry the dynamics of the temperature field of a number of PAA hydrogels and a cartilage tissue exposed to laser radiation at 1.56 μm. Based on the radiometric measurements of temperature and theoretical calculations of the temperature field dynamics, we estimated the thermal diffusivity, specific heat, and effective absorption coefficient and determined the composition of the PAA hydrogel having thermal and optical properties close to those of a cartilage tissue.

## 2. Materials and methods

Biological tissue samples were prepared from the hyaline cartilage of a calf nasal septum. Cartilage samples were prepared on the day of experiments and were kept in a physiological solution at 4 °C to minimise the destruction of the tissue caused by natural decomposition processes [15]. The cartilage was cleaned from perihondrium and mucosa immediately before experiments and 1.5-mm-thick discs of diameter 10 mm were obtained from it.

Polyacrylamide hydrogels were synthesised by radical copolymerisation of acrylamide (Lancaster) and N,N′-methylenebisacrylamide (Amresco) in the presence of catalytic amounts of ammonium persulfate (Amresco) and N,N,N,N′-tetramethylethylenediamine (Lancaster) [16 – 18]. Acrylamide and N,N′-methylenebisacrylamide were

**A.V. Kondyurin** Department of Chemistry, M.V. Lomonosov Moscow State University, Vorob'evy gory, 119992 Moscow, Russia; e-mail: andrey.kondyurin@gmail.com;
**A.P. Sviridov** Institute of Laser and Information Technologies, Russian Academy of Sciences, ul. Svyatoozerskaya 1, 140700 Shatura, Moscow region, Russia; e-mail: sviridov@laser.ru

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dissolved in distilled water and the solution was poured into a cylindrical vessel of diameter 10 mm. Then a catalytic amount of $(NH_4)_2S_2O_8$ and N,N,N′,N′-tetramethyethylenediamine was added. After termination of the reaction, the vessel was cooled and hermetically sealed to prevent the evaporation of water.

We synthesised four groups of PAA gels with degrees of cross-linking 1 : 9, 1 : 14, 1 : 19, and 1 : 24. The water mass content in PAA gels of each group was 90 %, 85 %, 80 %, 75 %, and 70 %. The choice of the concentration interval of PAA gels was limited, on the one hand, by the limiting solubility of initial reagents in water, which prevented the preparation of gels with a lower water content, an on the other hand, by the limiting concentration of initial reagent in water below which gels could not be synthesised. The degree of cross-linking of gels was determined by the ratio of the initial amounts of N,N′-methylenebisacrylamide and acrylamide.

The thermal analysis of samples was performed by the method of differential scanning calorimetry by using a DSC 30 scanning calorimeter (Mettler TA 4000, Switzerland). The analysis showed that the thermal behaviour of PAA gels and the cartilage tissue in the temperature range from 25 to 60 °C was the same (up to the denaturation temperature of the hyaline cartilage) and any considerable power-consuming processes, which could distort the temperature field upon laser heating of samples, were absent in gels.

The temperature fields induced by the 1.56-μm laser radiation in PAA gels and the cartilage tissue were investigated by the method of IR radiometry. Samples were mounted vertically in a special holder so that both sides of a disc sample were in contact with air. One side of the sample was irradiated for 1 s by a 0.5-W LS-1,56-5 erbium fibre laser (IRE-Polyus Scientific and Technical Association). The laser radiation was delivered through a silica fibre of diameter 0.6 mm. The distance from the fibre end to the sample surface was 7 mm. The scanning of the laser radiation intensity distribution with a small-aperture sensor showed that it was well described by the Gaussian distribution at different distances from the fibre end. The temperature field was detected on the unilluminated side of the sample along a line passing through the laser beam centre with the help of an IRTIS-200 infrared imager (Russia) with an accuracy of 0.05 °C and a spatial resolution of 0.1 mm. The frame repetition rate was 70 Hz. The laser-induced temperature field was recorded for 20 s both during sample heating and cooling. The increase in the sample temperature upon heating did not exceed 6 °C.

The thermal properties of PAA gels were determined by comparing the measured temperature field with the model field. By varying the values of required parameters, namely, the thermal diffusivity, the effective attenuation of the laser beam and the heat capacity of a sample by using the modified Levenberg–Marquardt algorithm [19], we obtained the minimal discrepancy between the calculated and measured temperature fields, which was a criterion for the choice of the corresponding values of these parameters.

The method that we used is in fact analogous to that proposed in papers [20, 21], where the temperature field was produced by 1-μs pulses from a $CO_2$ laser, which were absorbed in a thin surface layer ($\sim 20$ μm) of biological tissues and phantoms containing water. This allowed the authors to separate spatial variables and to find the analytic solution of the heat conduction equation describing the temperature dynamics of a surface layer during cooling. In our case, the approach used in [20, 21] cannot be applied because radiation is absorbed in a rather thick layer ($\sim 1.5 - 2$ mm) during a long time (1 s). Because of this, we solved the heat conduction equation numerically by the method of finite differences. In addition, we used the numerical solution of the heat conduction equation including both cooling and heating stages. This allowed us to estimate not only the thermal diffusivity, as in papers [20, 21], but also the specific heat and the effective attenuation of laser radiation.

The three-dimensional temperature field induced by IR laser radiation was calculated by using the classical heat conduction equation in cylindrical coordinates

$$\frac{\partial T}{\partial t} = \frac{1}{r}\frac{\partial}{\partial r}\left(\chi r \frac{\partial T}{\partial r}\right) + \frac{\partial}{\partial z}\left(\chi \frac{\partial T}{\partial z}\right) + f(r,z,t), \quad (1)$$

where $\chi(r,z)$ is the thermal diffusivity (in $\mathrm{mm^2\ s^{-1}}$); $T(r,z,t)$ is temperature at the point with coordinates $r$ and $z$ at the instant $t$; and $f(r,z,t)$ is the heating rate of the medium caused by the absorption of laser radiation (in $\mathrm{K\ s^{-1}}$). It is assumed that the radiation intensity inside a sample is described by the Beer law. Then, we have for Gaussian light beams

$$f(r,z,t) = (1 - R_\mathrm{d})\frac{\alpha E_0}{C_p \rho}\exp\left[-2\left(\frac{r}{w_\mathrm{L}}\right)^2\right]\exp(-\alpha z)\theta(t), \quad (2)$$

$$\theta(t) = \begin{cases} 1 \text{ for } 0 < t < t_\mathrm{p}, \\ 0 \text{ for } t > t_\mathrm{p}, \end{cases}$$

where $\alpha$ is the effective attenuation of laser radiation (in $\mathrm{cm^{-1}}$); $C_p\rho$ is the specific heat per unit volume (in $\mathrm{J\ cm^{-3}\ K^{-1}}$) (hereafter, simply called specific heat); $\rho$ is the density (in $\mathrm{g\ cm^{-3}}$); $t_\mathrm{p}$ is the laser pulse duration (in nanoseconds); $E_0$ is the laser radiation intensity (in $\mathrm{W\ cm^{-2}}$); $w_\mathrm{L}$ is the laser beam radius (in mm); $R_\mathrm{d}$ is the diffuse reflection factor equal to 0.10, which was measured earlier by using an integrating sphere. In reality the spatiotemporal distribution of the laser radiation power density depends on the absorption and scattering coefficients and the scattering anisotropy of an optically inhomogeneous medium and strictly speaking is not described by the Beer law. To find this distribution, it is necessary to solve the problem of the propagation of light in such media, for example, by the Monte-Carlo method. Nevertheless, the use of the exponential law with some generalised coefficient $\alpha$ for the description of the power distribution of thermal sources in an optically inhomogeneous absorbing medium seems quite justified [22]. We will call this coefficient the effective absorption coefficient.

The temperature distribution in a sample at the initial instant is homogeneous and its temperature is equal to the temperature $T_0$ of the environment:

$$T(r,z,t=0) = T_0. \quad (3)$$

The conditions specified on the sample boundary are presented in Table 1. The value of $h$ was taken from [23]. Note that the influence of this parameter on the required quantities is insignificant. Thus, the twofold increase or decrease in $h$ causes the change in the required quantities only by 2 % – 3 %.

**Table 1.** Boundary conditions used in the formulation of the problem.

| Boundary | Boundary condition |
|---|---|
| $r = 0$ | $\lim_{r\to 0}\left(r\chi\frac{\partial T}{\partial r}\right) = 0$ |
| $r = R$ | $-\chi\frac{\partial T(r,z,t)}{\partial r}\Big\|_{r=R} = \beta T(R,z,t) - \mu$ |
| $z = 0$ | $-\chi\frac{\partial T(r,z,t)}{\partial z}\Big\|_{z=0} = \beta T(r,0,t) - \mu$ |
| $z = Z$ | $-\chi\frac{\partial T(r,z,t)}{\partial z}\Big\|_{z=Z} = \beta T(r,Z,t) - \mu$ |

Note: $\beta = h/C_p \geqslant 0$; $\mu = \beta T_0$; $h$ is the convective heat transfer coefficient with the environment (in W mm$^{-2}$K$^{-1}$); $R$ and $Z$ are the radius and thickness of a sample, respectively.

The formulation of the problem assumes that the thermal diffusivity, effective absorption coefficient, and specific heat are independent of temperature, coordinates, and time.

We found the required quantities $\chi$, $\alpha$, and $C_p\rho$ by minimising the functional

$$F(\chi,\alpha,C_p\rho) = \sum_{j=1}^{M}\sum_{i=1}^{N}[R_{\text{exp}}(r=r_j,t_i) - R_{\text{calc}}(r=r_j,t_i)]^2 \quad (4)$$

with the help of the modified Levenberg–Marquardt algorithm [19]. Here, $R_{\text{exp}}(r=r_j,t_i)$ is the radiometric temperature at the point $r_j$ at the instant $t_i$ and $R_{\text{calc}}(r=r_j,t_i)$ is the radiometric temperature at the point $r_j$ calculated by solving numerically the heat conduction equation. The temperature field dynamics for each sample was measured five times, which allowed us to calculate five statistically independent groups of thermal parameters $\chi$, $\alpha$, and $C_p\rho$, their average values and standard deviations. The relative measurement errors for $\chi$, $\alpha$ and $C_p\rho$ were no more than 3 %, 5 %, and 5 %, respectively.

## 3. Results and discussion

Radiometric measurements gave the sequence of one-dimensional temperature distributions on the unilluminated surface of the sample along a line passing through the maximum of the temperature field (Fig. 1a). The position of this maximum coincides in fact with the centre of the laser-beam section by the sample surface. The sample temperature increased during the laser pulse and continued to increase after laser switching off. The temperature rise time depended on the distance to the laser beam centre. The spatial distribution of the temperature field at the initial stage is specified by the function $f(r,z,t)$ and is mainly determined by parameters $\alpha$ and $C_p\rho$. The influence of the thermal diffusivity $\chi$ at this stage proves to be insignificant.

After irradiation, the temperature field is gradually redistributed due to the thermal flow directed inside the sample and along its surface and also due to convective heat exchange with the environment. In the given case, the temperature field dynamics was mainly determined by the thermal diffusivity $\chi$.

Thus, it is advantageous to calculate parameters $\chi$, $\alpha$ and $C_p\rho$ by using the spatiotemporal temperature distribution including both sample cooling and heating stages. It was sufficient to use in calculations the time dependences of temperature only for ten points with different coordinates, which are characterised by a small relative error of measuring temperature at them (Fig. 1b) during the time $t = 7$ s after the laser pulse onset. An increase in the number of coordinate points did not improve noticeably the measurement accuracy, and for $t > 7$ s the effect of noise in the measurements of the temperature of the cooled sample became noticeable.

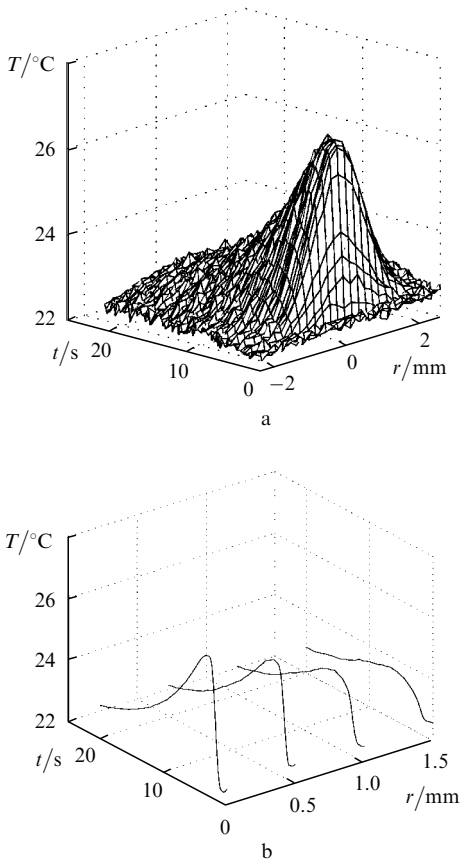


**Figure 1.** Temperature field dynamics on a sample surface (a) and time dependences of temperature used for calculating the thermal diffusivity, specific heat, and effective absorption coefficient (b).

The thermal diffusivity, specific heat, and effective absorption coefficient obtained for the nasal septum cartilage were $0.130 \pm 0.006$ mm$^2$ s$^{-1}$, $3.5 \pm 0.2$ J cm$^{-3}$ K$^{-1}$ and $10.7 \pm 0.7$ cm$^{-1}$, respectively. Note that these values of the thermal diffusivity and specific heat well agree with the values $\chi = 0.128 \pm 0.003$ mm$^2$ s$^{-1}$ and $C_p\rho = 3.7 \pm 0.3$ J cm$^{-3}$ K$^{-1}$ obtained in [24]. The value of the effective absorption coefficient also seems quite reasonable because the absorption coefficient of water at 1.56 μm is $\sim 10.5$ cm$^{-1}$ [25]. The direct calculation of the temperature filed dynamics by using cartilage parameters presented above well agrees with the temperature dynamics both at the centre and periphery of the irradiated region (Fig. 2).

Table 2 presents the values of the thermal diffusivity, specific heat, and effective absorption coefficient measured for PAA hydrogels with different concentrations and different degrees of cross-linking. Note that the values of $\chi$, $C_p\rho$ and $\alpha$ vary approximately in intervals 0.12–0.19 mm$^2$ s$^{-1}$,

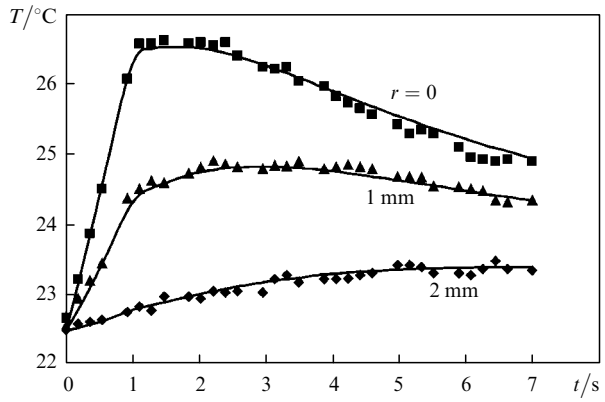


**Figure 2.** Dynamics of the temperature field induced by 1-s, 0.5-W laser pulses on the surface of a hyaline cartilage at different distances $r$ from the laser beam centre. Points are experiment, curves are calculations.

$2.5-5.5$ J cm$^{-3}$ K$^{-1}$, and $9-17$ cm$^{-1}$, respectively. One can see that the corresponding parameters for cartilage also lie in these intervals. The PAA hydrogel with the degree of cross-linking 1 : 9 containing 70 % of water has parameters $\chi = 0.139 \pm 0.004$ mm$^2$ s$^{-1}$, $C_p\rho = 3.6 \pm 0.1$ J cm$^{-3}$ K$^{-1}$, and $\alpha = 10.8 \pm 0.5$ cm$^{-1}$ which are most close to those of the nasal septum cartilage. Spatiotemporal temperature distributions on the surfaces of hydrogel and cartilage samples were also virtually identical.

One can see from Table 2 that, by varying the degree of cross-linking and water content in PAA gels within the limits indicated above, we can realise numerous different combinations of parameters $\chi$, $C_p\rho$ and $\alpha$. However, the behaviour of these parameters depending on the degree of cross-linking and water content is nonmonotonic in a number of cases. Thus, a gel with the degree of cross-linking 1 : 9 containing 80 % of water has the minimal thermal diffusivity (Fig. 3). Therefore, some gels with different degrees of cross-linking and (or) different water contents can have the same values of some of the parameters $\chi$, $C_p\rho$ and $\alpha$. However, not all combinations of these parameters can be realised within the above-mentioned limits because three parameters should coincide, while only two parameters are varied (the degree of cross-linking and water content). An important criterion in the choice of the thermal optical equivalent of a biological tissue is the relation between the water content and organic component of a PAA gel, which should be as close as possible to this relation in the biological tissue. According to all the properties listed above, the PAA gel with the degree of cross-linking containing 70 % of water appears the most convenient for simulating laser-induced temperature fields.

**Table 2.** Thermal diffusivity, specific heat, and effective absorption coefficient at 1.56 μm measured with the accuracy of ±5 % for PAA hydrogels with different degrees of cross-linking and different water contents.

| Degree of cross-linking | Water content (%) | $\chi$/ mm$^2$ s$^{-1}$ | $C_p\rho$/J cm$^{-3}$ K$^{-1}$ | $\alpha$/cm$^{-1}$ |
|---|---|---|---|---|
| 1 : 9 | 90 | 0.134 | 4.5 | 12.7 |
| | 85 | 0.124 | 4.9 | 16.1 |
| | 80 | 0.118 | 5.1 | 17.4 |
| | 75 | 0.158 | 3.9 | 12.2 |
| | 70 | 0.139 | 3.6 | 10.8 |
| 1 : 14 | 90 | 0.193 | 2.6 | 11.6 |
| | 85 | 0.152 | 3.7 | 12.4 |
| | 80 | 0.150 | 3.9 | 13.7 |
| | 75 | 0.142 | 4.8 | 13.3 |
| | 70 | 0.149 | 3.0 | 9.1 |
| 1 : 19 | 90 | 0.165 | 3.6 | 13.0 |
| | 85 | 0.151 | 3.7 | 12.4 |
| | 80 | 0.149 | 3.8 | 12.0 |
| | 75 | 0.159 | 3.4 | 10.8 |
| | 70 | 0.155 | 3.6 | 11.4 |
| 1 : 24 | 90 | 0.151 | 4.0 | 12.8 |
| | 85 | 0.156 | 3.4 | 11.2 |
| | 80 | 0.122 | 5.5 | 12.8 |
| | 75 | 0.149 | 3.8 | 11.5 |
| | 70 | 0.162 | 3.1 | 10.4 |

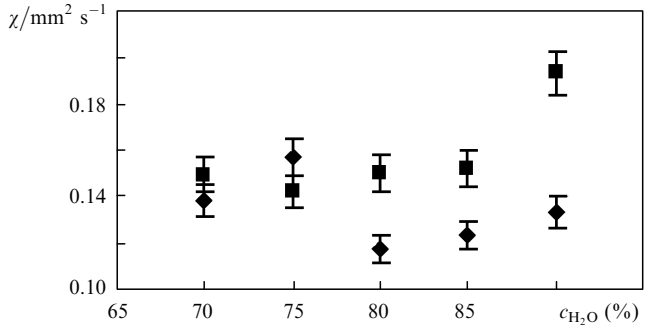


**Figure 3.** Dependence of the thermal diffusivity $\chi$ of PAA gels on the water content $c_{H_2O}$ for degrees of cross-linking 1 : 9 (◆) and 1 : 14 (■).

Figure 3 presents the dependences of the thermal diffusivity for a number of PAA hydrogels with different water contents and different degrees of cross-inking. Note that the dependence of the heat-conducting properties of the PAA hydrogel on the water content proved to be nonmonotonic and the heat-conducting properties of PAA hydrogels containing 70 % and 75 % of water are almost independent of the degree of cross-linking of a polymer network. For PAA hydrogels with a larger water content, a decrease in the degree of cross-linking leads to a considerable increase in the thermal diffusivity.

The dependence of the effective coefficient of absorption of laser radiation by PAA hydrogels with degrees of cross-linking 1 : 9 and 1 : 14 on the water content has a maximum (Fig. 4a). This maximum is most pronounced for PAA hydrogels with the maximum degree of cross-linking (1 : 9). The dependence of $\alpha$ on the water content for PAA hydrogels with degrees of cross-linking 1 : 19 and 1 : 24 becomes monotonic (Fig. 4b). The effective absorption coefficient (at 1.56 μm) of inhomogeneous media depends on the absorption coefficient $\mu_a$, scattering coefficient $\mu_s$, and anisotropy factor $g$ and, according to the diffuse model of light propagation, is described by the expression

$$\alpha \approx \sqrt{3\mu_a[\mu_a + \mu_s(1-g)]}. \quad (5)$$

The main chromophore absorbing IR radiation at 1.56 μm in these systems is water, and we can assert that an increase

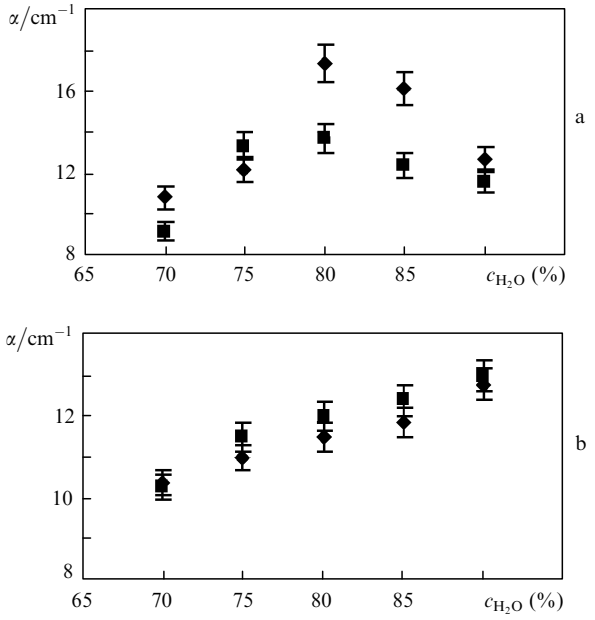


**Figure 4.** Dependences of the effective absorption coefficient $\alpha$ at 1.56 μm for PAA gels on the water content $c_{H_2O}$ for degrees of cross-linking 1 : 9 (◆), 1 : 14 (■) (a) and 1 : 19 (◆) and 1 : 24 (■) (b).

in its content in a PAA hydrogel leads to a linear increase in the coefficient $\mu_a$.

The dependence of $\mu_s$ on the hydrogel concentration is considerably more complicated. It is known that upon copolymerisation numerous statistical hydrogel balls are formed in which PAA chains are disordered [18]. These balls grow during polymerisation, come close together to form transverse cross-links under the condition that the polymer chain is embedded with a sufficient number of polyfunctional monomers of the N,N′-methylenebisacrylamide type, which can react with free end functional groups of other chains. The polymer concentration and the number of transverse cross-links in the PAA hydrogel determine the number of scattering centres. If the structure, shape, and size of these centres are constant and the distance between them is large enough, the scattering coefficient of the medium will be proportional to their concentration. In the case of PAA hydrogels, these conditions are probably not always fulfilled, and $\mu_s$ depends on the water content and the number of cross-links in a complicated way. Unfortunately, at present there are no adequate models describing light scattering in PAA hydrogels.

The equivalent of a cartilage tissue proposed in our paper was used to calibrate a system for controlling laser operations on nasal septum shaping [26]. These operations are performed by using a laser instrument with a sapphire window through which the 1.56-μm laser radiation is directed on the nasal septum. The sapphire window of the instrument is pressed against the nasal septum surface, two thermocouples symmetrically mounted at the edges of the sapphire window controlling the temperature of mucosa. However, it was necessary first to compare the readings of thermocouples, measuring temperature at the sapphire window periphery, with temperature in the internal region of the nasal septum at which the therapeutic effect is achieved. For this purpose we used the PAA hydrogel with the degree of cross-linking 1 : 9 and water content 70 %. The gel was polymerised in special cylindrical organic glass cells inside which thermocouples were placed (Fig. 5). The sapphire window of the laser instrument was pressed against the PAA gel surface, the window being located a distance of 0.5 mm below the thermocouple. When a laser was switched on, signals from the thermocouple located in the cell with the gel and thermocouples mounted on the laser instrument were simultaneously detected. By assuming that temperature fields in the cartilage and its PAA equivalent are identical, we have managed to relate the readings of instrument thermocouples with the real temperature inside the nasal septum during operations.

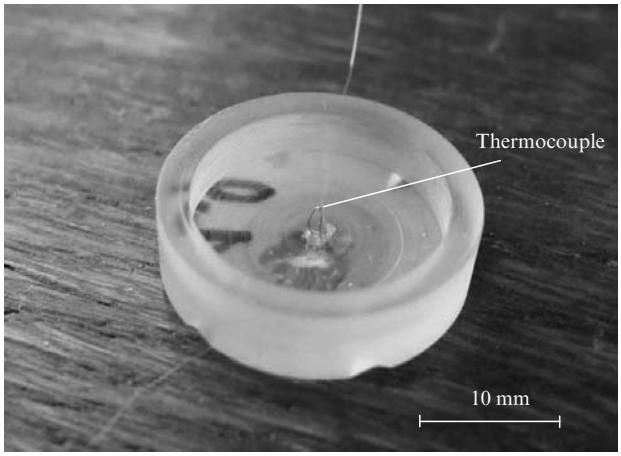


**Figure 5.** Cell for calibrating a laser instrument.

## 4. Conclusions

We have measured by the method of IR laser radiometry the thermal diffusivity, specific heat, and effective absorption coefficient at 1.56 μm for cartilages and a number of PAA gels with different water contents and different degrees of cross-linking. These parameters for the PAA gel with the degree of cross-linking 1 : 9 containing 70 % of water and the nasal septum cartilage proved to be close. The equivalent of cartilages obtained in our study can be used to simulate laser-induced temperature fields.

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