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Change in the optical properties of hyaline cartilage heated by the near-IR laser radiation

V N Bagratashvili, N V Bagratashvili, V P Gapontsev, G Sh Makhmutova, V P Minaev, A I Omel'chenko, I E Samartsev, A P Sviridov, E N Sobol', S I Tsypina

***Abstract*. The *in vitro* dynamics of the change in optical properties of hyaline cartilage heated by fibre lasers at wavelengths 0.97 and 1.56 μm is studied. The laser-induced bleaching (at 1.56 μm) and darkening (at 0.97 μm) of the cartilage, caused by the heating and transport of water as well as by a change in the cartilage matrix, were observed and studied. These effects should be taken into account while estimating the depth of heating of the tissue. The investigated dynamics of light scattering in the cartilage allows one to choose the optimum radiation dose for laser plastic surgery of cartilage tissues.**

Keywords: *fibre laser, cartilage, absorption and scattering of radiation, stress relaxation.*

1. Introduction

The phenomenon of IR laser-induced relaxation of stresses in the cartilage tissue, which was discovered and studied in papers [1 – 4], seems to be quite promising for developing an entirely new type of medical operations and procedures – a controlled reshaping of cartilage tissues. One of these applications is the reshaping of the cartilage of the deformed nasal septum of a human being (septonasal correction), which is already at the stage of implementation in clinical practice [5, 6].

The biochemical and cell responses of the cartilage tissue to laser heating are quite sensitive to the temperature distribution in space and time. The heating of the cartilage tissue exposed to near-IR laser radiation is mainly caused by the absorption of radiation by water molecules [7]. Note that absorption and scattering of IR radiation may vary significantly during laser heating, i. e., the heating problem becomes essentially nonlinear. This is due to the microscopic and macroscopic processes proceeding in the biological tissue upon its heating. The microscopic processes include, in particular, the deaggregation of water and the transition of bond water to the 'free' state (the fraction of bond water in the cartilage is about of 4 % [8]). This transition occurs in the cartilage tissue at a temperature about of 70°C and is accompanied by an increase in the mobility and, probably, the liberation of proteoglycanes in the cartilage matrix. The macroscopic processes include the heat- and mass transport (of water) from the laser-heated cartilage region. Obviously, all these processes can also change light scattering in the cartilage [9, 10].

Various types of lasers were used in the experiments on the cartilage tissue reshaping. It was shown [11] that irradiation of a 1-mm thick cartilage by a CO_2 laser leads to a strong overheating of the tissue surface followed by the cartilage destruction due to a very strong absorption ($\sim 10^3$ cm^{-1}) in the cartilage at this wavelength. It was shown earlier [2] that the cartilage reshaping without its destruction is possible only in a very narrow range of the cartilage temperature – laser-pulse duration diagram. A more uniform heating of the cartilage can be achieved by using a pulsed 2.09-μm holmium laser [2, 6, 12] or a 1.32-μm Nd : YAG laser [3]. However, a final choice of the most efficient laser for the cartilage plastic surgery has not been made so far, especially in the context of extensive clinical applications. Among the various lasers analysed by us for assessing their applicability for this purpose, near-IR diode lasers have attracted our attention. These compact, convenient, and high-power (up to tens of watts) lasers are exceptionally promising for various medical applications.

In this work, we study *in vitro* the optical properties of hyaline cartilage irradiated by a 0.97-μm diode laser and a 1.56-μm Er-doped fibre laser for determining the optimal conditions for the controlled cartilage reshaping.

2. Experimental

Fig. 1 shows the schematic of the setup for studying the optical and mechanical properties of cartilages exposed to laser radiation. Experiments were carried out on hyaline cartilages (nasal septum of four-month old pigs). The cartilage plate was cut in the form of rectangular bars of width 3 – 5 mm, length 10 – 12 mm and thickness between 0.5 and 2 mm.

A 0.97-μm diode laser with an optical fibre outlet (IRE – Polyus, model 'Lazon') and an output power up to 10 W, as well as a 1.56-μm Er-doped IR fibre laser (IRE – Polyus, model LS-1.56) with a maximum output power of 5 W [13],

V N Bagratashvili, N V Bagratashvili, A I Omel'chenko, A P Sviridov, E N Sobol', S I Tsypina Institute of Laser Problems and Information Technologies, Russian Academy of Sciences, Pionerskaya ul. 2, Troitsk, Moscow oblast, 142190 Russia
V P Gapontsev, V P Minaev Institute of Radio Engineering and Electronics – Polyus, pl. Acad. Vvedenskogo 1, 141120 Fryazino, Moscow oblast, Russia
G Sh Makhmutova Medical Centre, Management of Affairs of the President of the Russian Federation, Grokhol'skii per. 31, 129090 Moscow, Russia

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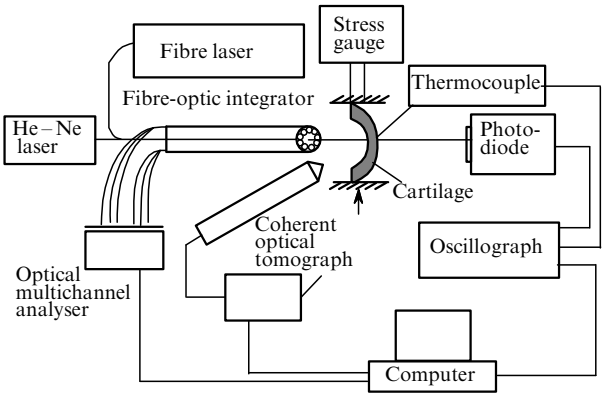

Figure 1. Schematic of the setup for studying the optical properties of cartilage tissue irradiated by laser.

were used as radiation sources. The size of the laser spot was varied from 1 to 10 mm by changing the distance between the fibre end and the cartilage surface.

The temperature inside the cartilage tissue was measured with the help of a needle thermocouple of diameter 30 μm inserted to a depth of 0.5 mm into a small cut made with a scalpel on the sample side opposite to the surface being exposed to laser radiation. The temperature dynamics of the irradiated surface of the cartilage was controlled with a fibre IR radiometer.

A calibrated strain gauge was used to record the change in the mechanical properties of the cartilage exposed to laser radiation (see Fig. 1). The gauge measured the force which the deformed cartilage exerts on it (bearing reaction). The internal stress was generated in the cartilage sample by moving the strain gauge [9].

The transmission of IR laser radiation through the cartilage sample for various levels of IR radiation was recorded by a germanium IR photodetector placed directly behind the sample. To exclude the deformation of the cartilage plate during laser irradiation and drying, the samples were placed in a special holder (not shown in Fig. 1) consisting of two metal plates with orifices for the laser beam. The output signals from the strain gauge, thermocouple, and IR detector were recorded on an oscillograph and a computer.

A specially designed fibre-optic system was used to measure the angular distribution of the visible light at 0.632 μm scattered backward from the cartilage sample (see Fig. 1). The system consists of a set of optical fibres whose ends form two concentric rings and are located directly in front of the surface of the exposed sample. The probe radiation from a 0.632-μm He – Ne laser and radiation from a fibre laser are directed through fibres located at the centre of the system. The He – Ne laser radiation scattered in the sample was detected by the optical multichannel analyser during irradiation of the sample by the IR laser.

3. Results and discussion

3.1 Dynamics of IR transmission

To provide the optimal conditions during irradiation of cartilage by laser aimed at its reshaping, it is necessary that the penetration depth of radiation into the tissue be of the order of the thickness of the material being processed. The absorption of the cartilage tissue in the IR region is mainly determined by the water contained in it. For example, the absorption coefficient of 'free' water at 1.56 μm is ~ 10 cm^{-1}. Because the concentration of water in freshly prepared cartilage is about of 70 % – 80 %, the penetration depth for laser radiation at such a wavelength into the cartilage tissue is about of 1.2 – 1.3 mm. Our direct measurements of transmission of thin cartilage samples at very low intensities of the laser beam ($P = 0.01$ W) give a slightly smaller penetration depth (1 mm) which may be due to a smaller contribution of scattering to the overall transmission (the scattering coefficient of 1.56-μm radiation in the cartilage is more than an order of magnitude lower than the absorption coefficient). On the contrary, the scattering coefficient at 0.97 μm in the cartilage (5 – 10 cm^{-1}) is much higher than the absorption coefficient (0.35 cm^{-1}).

Our experimental studies showed that the dynamics of transmission of IR laser radiation through a cartilage plate was different at the wavelengths 0.97 and 1.56 μm. Fig. 2a shows the time dependences of transmission of laser radiation at 1.56 μm through cartilage samples for various input powers of the laser. One can see that these dependences are determined by the laser radiation intensity. For example, the transmission through the cartilage does not change with time during low-intensity irradiation. However, transmission begins to increase with the exposure time at higher laser powers. In other words, the tissue is bleached upon irradiation at a given wavelength. The higher the radiation intensity, the stronger the bleaching of the tissue. For the highest radiation power shown in Fig. 2a, the transmission through the sample after 7 seconds of irradiation increased by a factor of about 2.5 relative to its initial value.

The dynamics of transmission of laser radiation at 0.97 μm also depends on the laser radiation power (Fig. 2b). The transmission remains virtually unchanged for power $P < 5$

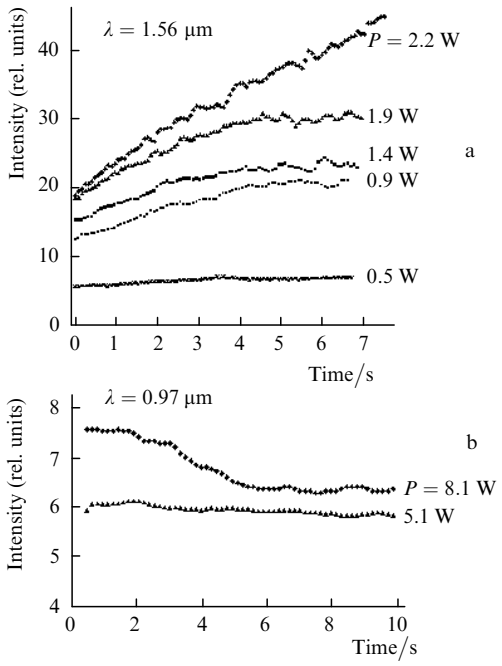

Figure 2. Time dependences of the intensity of radiation at 1.56 (a) and 0.97 μm (b) transmitted through the cartilage sample at different output powers P of lasers (cartilage thickness is 1.7 mm, spot size at the sample is 3 mm).

W, which means that the tissue heating under these conditions is apparently insignificant. However, a clear decrease in the transmission is observed for $P > 5$ W, i.e., a darkening of the tissue takes place.

In our opinion, the observed difference in the dynamics of IR transmission through a cartilage plate at 0.97 and 1.56 μm is due to different contributions of absorption and scattering, as well as their temperature dependences, to the transmission. The self-induced bleaching of the cartilage irradiated at 1.56 μm is caused mainly by heating of the water contained in it. First, the heating causes a deaggregation of the water molecules, thereby decreasing the oscillator strength of the water molecule in the IR absorption, as well as a shift of the absorption band [14]. Second, the emerging temperature gradient stimulates the transport of water from the irradiated region of the sample to its unexposed region. Moreover, heating accelerates the evaporation of water from the sample surface. For a 0.97-μm laser, the transmission is mainly determined by scattering, hence, an increase in scattering (see below) should lead to a decrease in transmission, as is indeed observed in the experiment.

3.2 Laser heating and stress relaxation

Relaxation of internal stresses in the cartilage tissue irradiated by a laser mainly depends on the tissue temperature and space–time distribution of this temperature. For a specific cartilage sample, this process is determined, first, by the laser radiation parameters (wavelength, intensity, dose) and, second, by the optical properties of the cartilage (IR absorption and scattering), as well as by the rate of heat transport from the region of irradiation.

In our experiments, we measured the cartilage tissue temperature using a thin thermocouple and a fibre IR radiometer. Note that such a thermocouple was not overheated upon irradiation by the laser beam and measured the temperature of the surrounding tissue. This follows directly from an analysis of the temperature dependences, and an identity of the results of measurements using a thermocouple and an IR radiometer (Fig. 3).

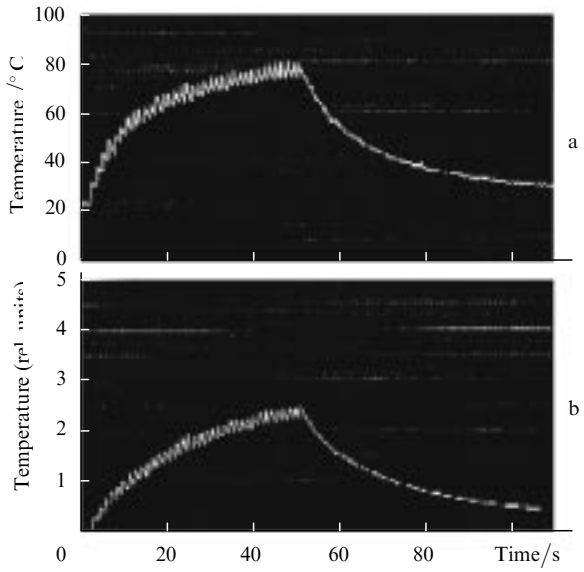

Figure 3. Oscillograms showing the variation of temperature in a cartilage sample irradiated by a repetitively pulsed laser: output signal of the thermocouple (a); output signal of the IR radiometer (b).

One can see from Fig. 3 that at the initial stage of irradiation ($0 < t \lesssim 10$ s), the sample temperature at the centre of the laser spot increases linearly with time, after which its increase becomes slower at about 60°C and it falls rapidly following the switching off of the laser. The departure from linearity at $t \gtrsim 10$ s is due to heat transport and temperature-induced decrease in the IR absorption by water (see Figs 2a and b). The relation between the duration of the laser pulse, stress in the cartilage, and its temperature is shown in Fig. 4.

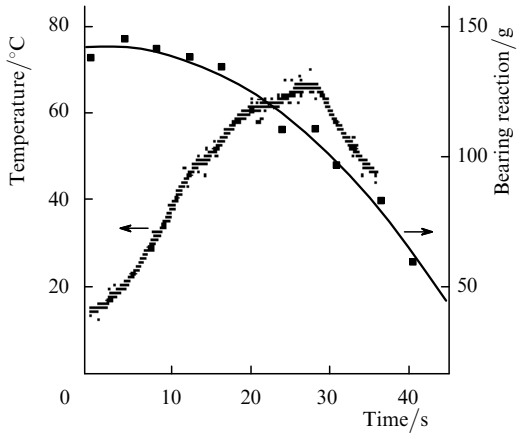

Figure 4. Temperature of the cartilage sample and stress relaxation upon its heating by laser radiation at 1.56 μm (sample thickness is 1.7 mm, $P = 2.7$ W, spot size on the sample is 5 mm).

At the very beginning of irradiation, the bearing reaction increases insignificantly due to the thermal expansion of the cartilage. Further irradiation leads to a gradual decrease in the bearing reaction. Our experiments show that the higher the laser intensity, the faster the decrease in the bearing reaction. A long-term stability of the new shape of the cartilage can be attained only by heating the tissue to 70°C and above. If the laser heating of the cartilage is terminated at a lower temperature, the initial shape of the cartilage immersed in physiological solution is completely restored after a certain period of time (from a few minutes to a few hours).

3.3 Dynamics of light scattering

To prevent heating of the cartilage by laser radiation, it is desirable to determine the radiation dose required for obtaining a stable relaxation of stresses (without the tissue damage) directly during sample irradiation, and to switch off the laser at the appropriate moment. The variations in the internal structure and mechanical properties of the tissue are accompanied by changes in its optical properties. Therefore, the intensity of the backscattered radiation of the probe laser can be used as the controlling parameter for the feedback system.

Experiments on the measurement of the intensity of the light scattered from the cartilage irradiated by a laser beam with the help of an integrating sphere demonstrated a direct correlation between the stress relaxation processes (or the change in the cartilage shape) and backscattering of light [9, 10]. However, it is impossible to use an integrating sphere under clinical conditions. Because of this, we developed and applied a quite simple fibre-optic system (see Fig. 1) for collecting the backscattered light from the cartilage sample irradiated by an IR laser.

Fig. 5 shows the time dependences of the intensity of backscattered light from the irradiated sample measured at various angles relative to the centre of the laser spot (curves *1–3*). These dependences differ from each other during irradiation and exhibit maxima at different instants of time. In addition, there is no correlation between their maxima and the cartilage temperature. However, the backscattered signal obtained by integrating over all the fibres of the optical system (curve *4*) achieves its maximum around 70°C, i.e., at a temperature when an irreversible stress relaxation occurs in the cartilage tissue. Such a difference in the behaviour of backscattered signals collected at various angles can be attributed naturally to the deformation of the cartilage surface during irradiation. This follows directly from the images of the laser-irradiated cartilage obtained with the help of a coherent optical tomograph (Fig. 6). One can see that for certain irradiation doses, a pit is formed as a result of evaporation of water.

Signals recorded with the help of our optical fibre system are the sum of the signal of light scattered in the bulk of the cartilage and that reflected from its surface (the surface reflection is clearly seen in the tomograms). Obviously, a change in the curvature of the surface alters the angular distribution of the intensity of light reflected from the surface. Spatial integration allows us to take this effect into account, and the curve for the integrated signal (curve *4* in Fig. 5) gives complete information on the temporal behaviour of backscattered radiation. Note that the similar dynamics of light scattering (initial growth of intensity followed by its decrease) was also observed upon heating of a hyaline cartilage by laser radiation at 0.97 μm.

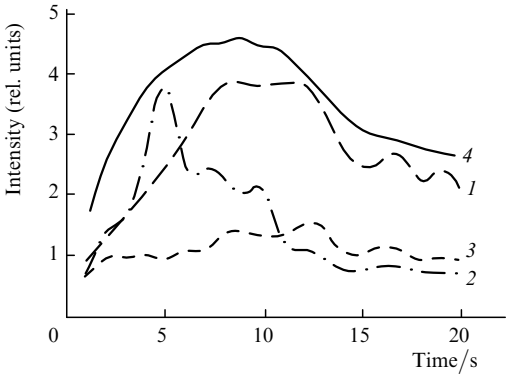

Figure 5. Time dependences of the intensity of light backscattered from a cartilage sample during heating by laser radiation (see text for notation).

One of the possible explanations for the observed nonmonotonic dynamics of light scattering consists in the following. The initial increase in the signal scattered from the sample being irradiated may be due to the escape of water from the irradiated region. Indeed, water and the cartilage matrix have different refractive indices, so that the removal of 'free' water in the tissue from the irradiated region obviously should enhance the intensity of the scattered signal. A further decrease in scattering observed at a temperature above 70°C is caused, in our opinion, by the liberation of bond water in the cartilage matrix [4, 9]. This leads to a decrease in the rigidity of the matrix, its compression, a decrease in the size of scattering centres and, consequently, to a decrease in the intensity of the scattered

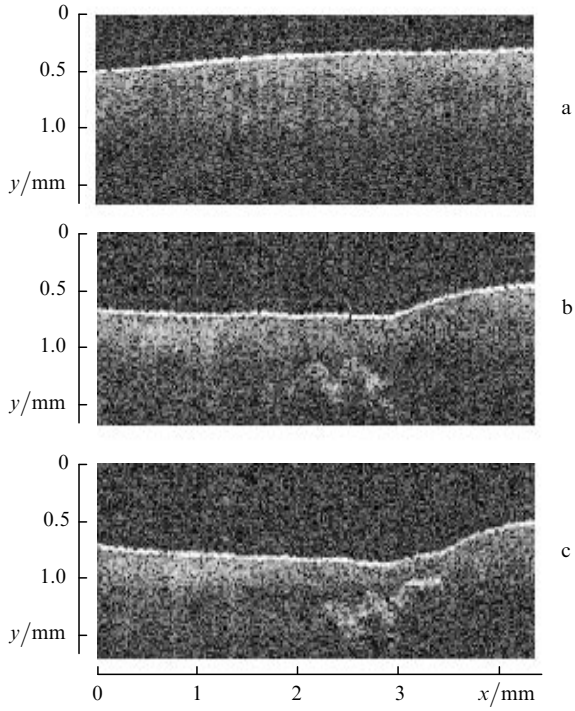

Figure 6. Images of the crosssection of the surface layer of a cartilage sample irradiated by a laser ($P = 2.2$ W, spot size is 4.6 mm), obtained from a coherent optical tomograph (a) at the initial instant of time; (b) after 34 s; and (c) after 52 s following the onset of irradiation.

light. The instant at which the intensity of the scattered light achieves its maximum should coincide with the onset of a decrease in the matrix rigidity and of stress relaxation, which is indeed observed in the experiments [15].

4. Conclusions

A moderate heating of the hyaline cartilage exposed to near IR laser radiation changes its optical properties.

The laser-induced bleaching (at 1.56 μm) or darkening (at 0.97 μm) of the cartilage is caused by a transport of the water inside the tissue and a change in the cartilage matrix. These effects should be taken into account while estimating the depth of tissue heating. A 1.56-μm fibre laser can be effectively used for a rapid uniform heating of thin cartilages (of thickness below 1–2 mm). For thicker cartilages (2–10 mm), a 0.97-μm laser can be used. Our investigations of the dynamics of light scattering in the cartilage at a temperature ensuring the relaxation of mechanical stresses (about 70°C) allow us to realise a simple fibre-optic control system for choosing the optimal irradiation dose in laser plastic surgery of the cartilage tissues.

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