Quantum Electronics **44** (1) 65–68 (2014)

BIOPHOTONICS

PACS numbers: 42.62.Be; 87.19.Pp; 87.50.W–; 87.64.Cc
DOI: 10.1070/QE2014v044n01ABEH015343

Optical properties of costal cartilage and their variation in the process of non-destructive action of laser radiation with the wavelength 1.56 μm

A.V. Yuzhakov, A.P. Sviridov, E.M. Shcherbakov, O.I. Baum, E.N. Sobol

Abstract. **The optical properties of costal cartilage and their variation under the action of laser radiation with the wavelength 1.56 μm are studied. The laser action regime corresponds to that used for changing the cartilage shape. The dynamics of the passed scattered laser radiation was studied by means of the optical fibre system, and the optical properties of the cartilage tissue (on the basis of Monte Carlo modelling of light propagation) – using the setup with two integrating spheres. Under the influence of radiation, the characteristics of which corresponded to those used for the cartilage shape correction, no essential changes in the optical parameters were found. The results obtained in the course of studying the dynamics of optical signals in the process of costal cartilage irradiation can be used for developing control systems, providing the safety and efficiency of laser medical technologies.**

Keywords: *laser, optical properties, cartilage tissue, shape correction.*

1. Introduction

The introduction of laser technologies into medicine contributes to the modern scientific progress. The modification of cartilage shape [1], particularly, the creation of implants from the costal cartilage for the treatment of larynx stenosis [2] is one of the examples of such technologies. To optimise the regimes of laser exposure and to provide the safety of laser medical technologies, especially when the rigorous control of the thermal field in the irradiated tissue is required [3], it is necessary to possess sufficiently precise knowledge of the tissue optical parameters and their variations in the process of laser exposure. Of particular interest are the studies of the dynamics of optical properties of the cartilage tissue under moderate heating with the radiation of the laser based on silica fibre doped with erbium ions with the wavelength 1.56 μm, which is now actively used in medical practice for correcting the shape of cartilage tissues and their regeneration [1, 4]. Such studies were carried out in Refs [5–13] for the cartilage of nasal septum and ear conch; however, for the costal cartilage the data on optical parameters are currently absent. Hence, the measurements of the costal cartilage optical parameters and their variation in the process of laser heating are urgent.

A.V. Yuzhakov, A.P. Sviridov, E.M. Shcherbakov, O.I. Baum, E.N. Sobol Institute on Laser and Information Technologies, Russian Academy of Sciences, ul. Pionerskaya 2, Troitsk, 142092 Moscow, Russia; e-mail: yuzhalvas@gmail.com, esobol@rambler.ru, baumolga@gmail.com

Received 12 November 2013
Kvantovaya Elektronika **44** (1) 65–68 (2014)
Translated by V.L. Derbov

The propagation of light trough an inhomogeneous medium can be unambiguously described if the absorption μ_a and scattering μ_s coefficients, the scattering anisotropy factor g, the refractive index and the scattering phase function are known [3, 14, 15]. In biomedical optics it is accepted that the scattering phase function, or the deflection angle probability density, is determined by the Henyey–Greenstein function. The calculations of μ_a, μ_s and g using the Monte Carlo method, based on the results of measuring diffuse reflection, diffuse transmission and their angular distributions, can lead to significant error in separating μ_s and g. This is because at the same value of the reduced scattering coefficient $\mu_s' = \mu_s \times (1 - g)$ the diffuse reflection, diffuse transmission, and their angular distributions weakly depend on the particular choice of μ_s and g. In many practical cases the thermal effect of laser radiation can be assessed with good accuracy using the effective absorption coefficient $\mu_{eff} = [3\mu_a(\mu_a + \mu_s')]^{1/2}$ [3]. In this connection the main attention in the present paper is paid to the measurement of the parameters μ_a and μ_s' of the costal cartilage and to their variations in the process of laser heating, simulating the medical procedures. Earlier analogous studies were performed by us in the eye tissues [16]. With the aim of developing a system for controlling the laser exposure process, the peculiarities of the signal behaviour are studied for diffuse reflection, diffuse transmission and paraxial transmission at moderate heating of the samples of costal cartilage with the radiation of medical laser having the wavelength 1.56 μm.

2. Materials and methods

Materials. The studies of optical properties were carried out using the samples of costal cartilage of a mature pig (the age 8–9 months). The fresh costal cartilage was stored up after the slaughter of animals and kept at the temperature –10 °C and humidity 100 %. Before the experiment the cartilage was defrosted and cut into plates with the thickness of 1.5–2 mm. From these plates by the aid of a punch the discs with the diameter 10 mm were cut, which were kept in saline to prevent drying. Before the experiment the samples were slightly dried at room temperature to avoid the absorption by the extra water at the tissue surface and fixed in a special holder. Each experiment was repeated at least four times.

Experimental setups. In the experiments we used the LC-1.56 fibre medical IR laser (IRE-Polus, Russia) generating the radiation with the wavelength 1.56 μm and the power up to 5 W, which allowed the control of laser pulse duration, interpulse separation and pulse repetition rate. Into the output optical fibre of the IR laser the probe cw laser radiation with the wavelength of 530 nm and the power 3–5 mW was introduced.

The IR radiation and the probe visible radiation transmitted through the sample were collected with the optical probe and sent to the multichannel optical analyser. The signals from the analyser were digitised and displayed on the PC monitor (Fig. 1). Since in the measurements using the optical fibre system a certain part of the radiation that does not hit the angular aperture of the collecting optical fibre is not taken into account, in order to obtain more precise information that takes the scattering of light in different directions into account, the experiments using doubled integrating spheres have been performed.

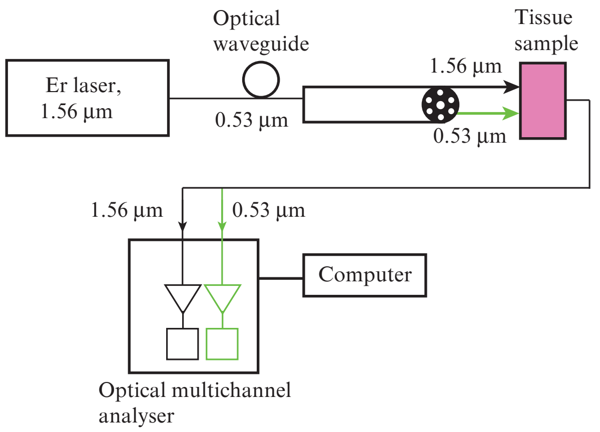

Figure 1. Scheme of the setup for studying optical properties of biotissues under laser exposure.

The setup (Fig. 2) consists of two spheres, the inner surfaces of which are coated with fine-dispersed barium sulphate that provides a high reflection coefficient for the radiation. As a result, a uniform light field is formed in the spheres, which does not depend on the direction of irradiation, and the signal can be detected at any point inside the spheres.

The IR laser radiation was supplied through the optical fibre (NA = 0.22, the core diameter 600 μm) normally to the sample surface. The regimes of laser exposure (the intensity ~70 W cm^{-2}, the pulse duration 500 ms, the pause duration 200 ms, the exposure time 6 s) corresponded to the conditions of laser shape correction, which were determined and studied

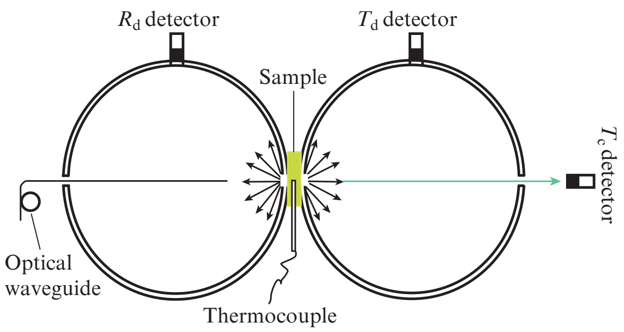

Figure 2. Scheme of the setup with two integrating spheres for measuring the optical parameters of biomaterials.

in Ref. [2]. Simultaneously, in the real-time regime the total diffuse reflection (detector R_d) was measured together with the diffuse (detector T_d) and paraxial (detector T_c) transmission, which were necessary for the Monte Carlo calculation of the optical parameters.

Calculation technique. The solution of the direct optical problem using the Monte Carlo method consists in multiple realisation of the photon random trajectory in the medium with the absorption and scattering taken into account [12]. The calculation of the photon trajectory is based on several rules, according to which the photon penetrates into the medium, propagates in it and leaves it. These rules are specified by the scattering and absorption probability that determines the free path length for the photon between the scattering and absorption events, the probability of the deflection of the photon trajectory by a certain angle, when the scattering occurs, and the probability of reflection of the photon from the boundary between the media with different refractive indices, when the photon attains such a boundary.

The calculation of the coefficients of diffuse reflection R_d, diffuse T_d and collimated T_c transmission was based on the updated Monte Carlo algorithm [17, 18]. In contrast to the traditional approach [19–21], in which the propagation of individual photons is simulated, in Ref. [17] it is proposed to consider a virtual photon package, propagating rectilinearly between the scattering events and losing its weight in the course of propagation in correspondence with the medium absorption coefficient. Such an approach significantly enhances the efficiency of modelling and reduces the time consumption for calculating R_d, T_d, and T_c at the given level of accuracy. In our case we considered 10^5 photon packages per iteration (the number of iterations was determined by the program in correspondence with the closeness of the initial parameters to the ones sought for). The accuracy of the measurement of μ_a and μ_s' was estimated as the variance of their values, calculated for 15 samplings from normal sets of measured values of R_d, T_d and T_c, with the measurement errors near the experimentally obtained values taken into account. The calculated measurement error amounted to 4%–8% depending on the experimental conditions and the parameter to be measured.

3. Results

The results of calculating the optical characteristics of costal cartilage based on the data, obtained with the use of two integrating spheres are as follows: before the exposure $\mu_a = 5.4 \pm 0.3$ cm^{-1}, $\mu_s' = 7.7 \pm 0.6$ cm^{-1}; after the exposure $\mu_a = 5.1 \pm 0.3$ cm^{-1}, $\mu_s' = 7.3 \pm 0.3$ cm^{-1}.

Within the measurement error there is no change in the light absorption and scattering coefficient in costal cartilage tissue exposed to laser radiation in the regimes, used to modify the cartilage shape.

The dynamics of optical signals in the process of laser action on the samples of costal cartilage is presented in Figs 3 and 4. At the initial stage of laser heating the active growth of signals occurs both for visible and IR ranges. Then at the same moments of time both curves in Figs 3 and 4 demonstrate a specific feature, expressed as the slowing of the growth. Further evolution of the IR signal demonstrates monotonic, although not so active, growth with attaining a plateau, while in the visible range the growing signal reaches an extremum followed by a fall. In Fig. 5 in the smaller time interval the synchronous character of the oscillations of the

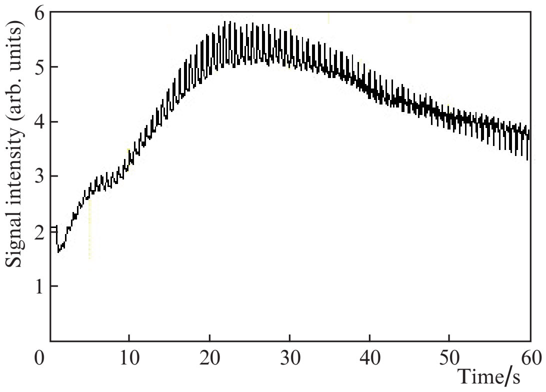

Figure 3. Dynamics of the transmitted visible radiation signals.

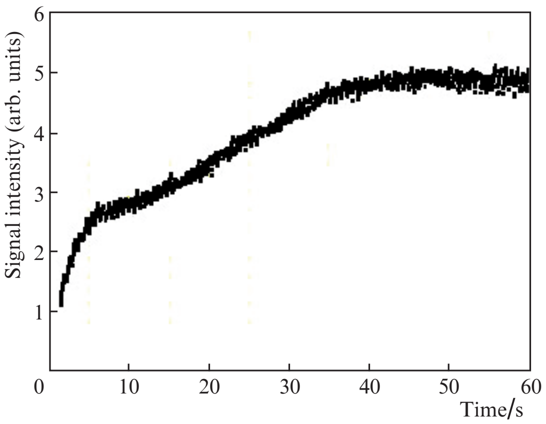

Figure 4. Dynamics of the transmitted IR radiation signals.

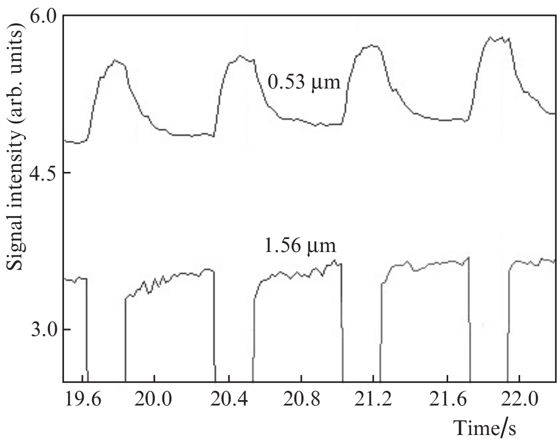

Figure 5. Dynamics of the signals of transmitted visible (0.53 μm) and IR (1.56 μm) radiation at the smaller time interval.

IR and visible light signals is shown, which demonstrates correspondence between the oscillation period and the period of pulsed laser action.

The dynamic curves presented in Figs 3–5 allow the following observations:

– the presence of specific features in both curves at the 5th and 6th second of laser action;

– the presence of clearly expressed extremum for the visible range in the interval 22–25 s;

– the absence of extremum for IR range and reaching the plateau by the signal after 35–40 s;

– the gradual growth of the visible radiation signal within the large temporal interval (up to 25 s) in spite of the decrease in the transmitted visible light intensity during each laser pulse with the duration 0.5 s and its growth in the pause between the laser pulses.

4. Discussion of the results

The data on the absence of essential changes in the optical parameters during the medical laser exposure agree with the earlier results on the absence of appreciable denaturation of the cartilage tissue [22] and the conservation of mechanical properties of the cartilage after its reshaping [2]. These data confirm the nondestructive character of laser correction of the costal cartilage shape.

The initial growth of the transmitted signal intensity both for visible and IR range (Figs 3 and 4) is most probably an evidence of redistribution of scattering centres due to the departure of water from the exposed zone. The simultaneous inflection of both curves is probably due to the deformation of the cartilage surface as a result of heating by the laser radiation. Further behaviour of the visible signal can be explained by the growth of the scattering anisotropy factor due to the increasing size of the scattering centres. The growth of the IR signal can be associated with the change in the water absorption spectrum with increasing temperature. The presence of a maximum and the following decrease in the transmission of the visible signal indicate the beginning of structural modifications, leading to the increase in the number of scattering centres. Reaching a plateau by the IR signal is an evidence of reaching the equilibrium between two oppositely acting processes, namely, the process of 'clearing' the cartilage because of the spectrum modification and the process of light scattering enhancement due to the production of additional interfaces between the media.

The synchronisation of the visible light signals with the period of the laser radiation pulsed regime is probably due to the appearance of gas bubbles under heating, their dissolving in the tissue and the tissue cooling during the pause between the pulses. The formation of gas bubbles under the laser heating is caused by the temperature dependence of solubility of gases that are always present in the interstitial fluid, which was earlier observed in articular and intervertebral cartilage [4, 23, 24]. Since in the cartilage tissues the surface of gas bubbles contains positively charged ions, such oscillations of gas bubbles can be accompanied also by oscillations of the cartilage tissue electric parameters in the course of pulsed-periodic laser exposure [25]. The complex kinetics of the signals in Figs 3–5 is related to the processes of formation, growth and collapse of gas bubbles in the irradiated tissue, which have different kinetics and different contribution to the system of scattering centres. The fall of intensity of transmitted visible light during each laser pulse with the duration 0.5 s (see Fig. 5) may be due to the formation of new gas bubbles, which leads to the increase in the number of scattering centres. The growth of intensity in the pause between the laser pulses may be caused by the collapse of some part of bubbles (i.e., the decrease of the number of scattering centres), and the gradual growth of the signal within a large time interval can be associated with the relatively slow increase in the mean size of gas bubbles due to the growth of the mean temperature of the tissue.

5. Conclusions

The absorption coefficient and the reduced scattering coefficient are measured in costal cartilage at room temperature under the laser heating in the regime of non-destructive reshaping. The absence of essential changes in optical parameters of the costal cartilage is demonstrated under laser heating in the medical regimes used for correcting the cartilage shape.

The dynamics of variation in the cartilage tissue transparency in the process of laser action can provide a background for recording desirable and undesirable changes of the cartilage structure in the designed control systems, providing the efficiency and safety of laser medical technologies.

Acknowledgements. The authors express their gratitude to A.I. Omelchenko for fruitful discussion.

The work was supported by the Russian Foundation for Basic Research (Grant Nos 13-02-01123, 13-02-00435, 12-02-31517, 12-02-31699).

References

1. Bagratashvili V.N., Sobol E.N., Shekhter A.B. (Eds) *Lazernaya inzheneriya khryashchey* (Laser Engineering of Cartilage) (Moscow: Fizmatlit, 2006).
2. Baum O., Soshnikova Yu., Sobol E., Korneychuk A., Obrezkova M., Svistushkin V., Timofeeva O., Lunin V. *Lasers in Surgery and Medicine*, **43**, 511 (2011).
3. Welch Ashley J., van Gemert Martin J.C. *Optical-Thermal Response of Laser-Irradiated Tissue* (Austin: Springer, 2011).
4. Sobol E., Zakharkina O., Baskov A., Shekhter A., Borschenko I., Guller A., Baskov V., Omelchenko A., Sviridov A. *Laser Phys.*, **19**, 825 (2009).
5. Sviridov A.P., Sobol E.N., Bagratashvilli N., Bagratashvili V.N., Omelchenko A., Dmitriev A.K., Shechter A., Ovchinnikov Y.M., Svistushkin V., Nikiforova G., Jones N., Lowe J. *Proc. SPIE Int. Soc. Opt. Eng.*, **2923**, 114 (1996).
6. Sobol E.N., Sviridov A.P., Kitai M.S., Gilligan J., Tolk N.H., Edwards G.S. *Proc. SPIE Int. Soc. Opt. Eng.*, **3601**, 122 (1999).
7. Bagratashvili N.V., Sviridov A.P., Sobol E.N., Kitai M.S. *Proc. SPIE Int. Soc. Opt. Eng.*, **3254**, 398 (1998).
8. Wong B.J., Milner T.E., Anvari B., Sviridov A.P., Omelchenko A.I., Bagratashvili V., Sobol E.N., Nelson J.S. *Proc. SPIE Int. Soc. Opt. Eng.*, **2970**, 380 (1997).
9. Bagratashvili V.N., Bagratashvili N.V., Gapontsev V.P., Makhmutova G.I., Minaev V.P., Omelchenko A.I., Samartsev I.E., Sviridov A.P., Sobol E.N., Tsypina S.I. *Kvantovaya Elektron.*, **31**, 534 (2001) [*Quantum Electron.*, **31**, 534 (2001)].
10. Sobol E.N., Sviridov A.P., Kitai M.S., Gilligan J., Edwards G.S. *Proc. SPIE Int. Soc. Opt. Eng.*, **3925**, 78 (2000).
11. Sobol E.N., Kitai M.S., Sviridov A.P., Butvina L.N. *Izv. Vyssh. Uchebn. Zaved., Ser. Prib.*, **46**, 47 (2001).
12. Sobol E.N., Kitai M.S., Sviridov A.P., Butvina L.N. *Izv. Vyssh. Uchebn. Zaved., Ser. Prib.*, **46**, 42 (2001).
13. Sviridov A., Kondyurin A. *J. Biomed. Opt.*, **15**, 055003 (2010).
14. Wang L., Jacques S.L. *Monte Carlo Modeling of Light Transport in Multi-layered Tissues in Standard C, Laser Biology Research Laboratory-17* (University of Texas, M.D.Anderson Cancer Center, 1992).
15. Lopatin V.V., Priezzhev A.V., Fedoseev V.V. *Biomed. Radioelektron.*, **7**, 29 (2000).
16. Yuzhakov A.V., Sviridov A.P., Baum O.I., Shcherbakov E.M., Sobol E.N. *J Biomed. Opt.*, **18** (5), 5800 (2013).
17. Wang L., Jacques S.L., Zheng L. *Comp. Methods Programs Biomed.*, **45**, 131 (1995).
18. Pope K., Wang L. *Proc. SPIE Int. Soc. Opt. Eng.*, **3914**, 300 (2000).
19. Roggan A., Albrecht H., Dorschel K., Minet O., Muller G. *Proc. SPIE Int. Soc. Opt. Eng.*, **2323**, 21 (1995).
20. Roggan A., Minet O., Schoder C., Muller G. *Proc. SPIE Int. Soc. Opt. Eng.*, **2100**, 42 (1994).
21. Roggan A., Schadel D., Netz U., Ritz J.-P., Germer C.-T., Muller G. *Appl. Phys. B*, **69**, 445 (1999).
22. Sobol E.N., Milner T.E., Shechter A.B., Baum O.I., Guller A.E., Ignatieva N.Y., Omelchenko A.I., Zakharkina O.L. *Laser Phys. Lett.*, **4** (7), 488 (2007).
23. Sobol E., Baskov A., Shekter A., Borshchenko I., Zakharkina O. *Light-Activated Tissue Regeneration and Therapy Conf.* (Springer Science + Business Media, LLC, 2008) Vol. 12, pp 259–266.
24. Omelchenko A.I., Sobol E.N. *Kvantovaya Elektron.*, **39** (3), 279 (2009) [*Quantum Electron.*, **39** (3), 279 (2009)].
25. Sobol E., Shekhter A., Guller A., Baum O., Baskov A. *J. Biomed. Opt.*, **16** (8), 80 (2011).