Optical coherence tomography monitoring of laser ablation processes in ENT tissues

M. PETRUS*, E.M. CARSTEA[a], D. C. A. DUTU, D. C. DUMITRAS
Department of Lasers, National Institute for Laser, Plasma and Radiation Physics,
409 Atomistilor St., PO Box MG-36, 077125, Bucharest, Romania
[a]National Institute of R&D Optoelectronics INOE 2000
409 Atomistilor St., PO Box MG-5, 077125 Bucharest, Romania

Laser ablation surgery diminishes the collateral symptoms and the recovery time. In this study our aims were to characterize tissue destruction after laser-ablation of the ENT (Ear, Nose and Throat) soft tissues. We compare the results obtained by Optical Coherence Tomography (OCT) method with conventional histology in order to evaluate *in vitro* the laser injury of the porcine larynges. For irradiation we have used the CO_2 laser surgical system. By adequate choice of the laser parameters, we managed to obtain a negligible damaged of the processed tissue due to the thermal effects. Using the heat transfer equation we have obtained a good agreement between the experimental and simulation data for the temperature distribution induced by different laser power in soft tissue.

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1. Introduction

In the recent years, laser technology has been introduced as an integral part of the therapeutic means in many centers of otolaryngology, head and neck surgery. In laser surgery pulse laser radiation is used to ablate the tissue and the goal is to minimize the collateral thermal damage in a desired region. The tissue at the aero-digestive tract is mainly formed by soft tissue, and is necessary a very precise cut, with a minimum thermal damage in the surrounded tissue. It has already been shown that benign and malign lesions of the glottis, the vocal cords, can be safely removed with the CO_2 laser rather than mechanically [1]. The vocal folds of the larynx permit the capacity for phonation. Structural changes on vocal folds affect the performance of phonation. Due to the expansion of lesions into deeper layers or tissue loss, the voice becomes unstable and will be limited in frequency and dynamic range. Moreover, scarring of the tissue caused by inadequate wound healing or disregard for the layered structure during surgery can implicate permanent disphonia [1].

To characterize the ENT soft tissue destruction after CO_2 laser ablation we use the optical coherence tomography (OCT) and histological analysis. OCT and histological estimates of the ablation crater dimensions and the depth of thermal injury were obtained. The CO_2 laser is the most optical source used in the ENT surgery, with a wavelength of 10.6 μm may be used for precisely cutting or vaporizing soft tissue with hemostasis. This long wavelength is strongly absorbed by water and has a penetration depth of about 10 μm into the tissue. Compared to the CO_2 laser, the application range of the other laser systems (diode laser, Er:YAG, Nd:YAG, Ho:YAG) is limited due to the specific characteristics of each type of laser radiation, such as limited cutting quality or the selective absorption of the laser light by pigments, especially hemoglobin, melanin and myoglobin. The main constituents of biological tissue which contribute towards absorption in the near infrared are water, fat and hemoglobin [2].

The objective of this study is to estimate the interaction effects of a CO_2 laser in soft tissue, the crater ablation depth and width (in relation with power and pulse duration). The purpose was to investigate experimentally and theoretically the thermal response of soft tissue irradiated by a laser source. The experimental results are compared with the solution of the heat equation in 1D model which demonstrates their applicability in determination the size of the crater ablation after laser pulse irradiation.

1.1 Optical Coherence Tomography method

Optical Coherence Tomography (OCT) is a novel non-invasive imaging technique based on low-coherence interferometry well established in ophthalmology and dermatology [3, 4, 5]. It is also well suited for imaging of the microstructures of the vocal cords. OCT technique is capable of producing cross-sectional images with an axial resolution of less than10 μm and a depth penetration of 1 to 2 mm in biological tissue. High sensitivity, large dynamic range, and micrometer level resolution imaging are achieved with an OCT technique by interferometric detection of backscattered light from the sample [6, 7, 8]. The OCT system used in our study is based on Fourier-Domain [9], Thorlabs OCP930SR Spectral Radar OCT device. The base unit contains a broadband super luminescent diode (SLD), a 930 nm light source, of 2 mW optical power, that is direct through an optical fiber couplet to the sample via the handheld scanning probe. The scanning device has a spectral bandwidth of 100 nm, 20 μm lateral resolution, 6.2 μm axial resolution and an image depth of approximately 1.6 mm. Images were acquires at 8 frames per second, an image size of 512 rows and the duration of the scan acquisition was 5 to 10 seconds [10].

1.2 Heat transport

When a tissue is exposed to the laser irradiation the tissue temperature increases due to the absorption process. Heat effects depend on the type of tissue and the temperature achieved inside the tissue [3, 7]. For example, the denaturation of the enzymes and the looseness of the membranes occurs at 40-45^{0}C; coagulation, necrosis and protein denaturation occurs at 60^{0}C; drying at 100^{0}C; carbonization occurs at 150^{0}C and pyrolysis and vaporization occurs at above 300^{0}C [11]. Temperature is the governing parameter of all thermal laser-tissue interactions. For the purpose of predicting the thermal response, a model for the temperature distribution inside the tissue must be derived.

Different types of lasers react differently with tissue and the interaction depends on: the wavelength of the laser, power density and exposure time, optical and thermal property of the tissue being irradiated, laser beam size on the tissue and if the exposure is CW or pulsed wave radiation.

The interaction mechanisms between the ENT soft tissue and the laser radiation depend on the time of the light exposure and the irradiance (the light energy delivered per unit area per unit time J/sec/cm^2 or the power per unit area, in W/cm^2). The spatial distribution of the ENT soft tissue damaged depends on the temperature distribution. The equation who describes the thermal diffusion, the temperature spread in the tissue is called the heat diffusion equation [12].

1.3 Modeling heat transport in soft tissue

The heat transfer in biological tissue irradiated is realized through four pathways: conduction, convection, evaporation and radiation [12,13,14]. For soft tissue the standard form of heat transport equation, where heat generation from metabolic activities is neglected [1, 9] is:

$$\rho C_p \frac{\partial T}{\partial t} = \nabla\left(k\nabla T\right) \tag{1}$$

where, ρ (kg/ m^3) is the tissue density, C_p (J /kg*K) is the specific heat capacity, k (W/ m*K) is called the thermal conductivity, T (K) is temperature.

With the implementation of the heat equation into Matlab we simulate the temperature distribution in irradiated soft tissue. We have obtained the temperature distribution along the laser propagation direction, i.e., the crater ablation depth. In order to simulate the tissue thermal response to laser radiation, one need the right values of the optical laser irradiation parameters e.g., laser power, beam diameter, wavelength, the laser beam profile, pulse duration.

2. Material and methods

The interaction process of tissue from the upper aero digestive tract (such as vocal cords) in the presence of laser radiations was investigated histological and with an optical tomography. As biological materials we used *in vitro* samples of porcine vocal cords.

The irradiation of tissue samples was made by a laser system, a CO_2 laser surgical system (SP-25LA, China), in pulse regime, the maximum output power for laser system being 25 W. For a good positioning precision and reproducibility, the laser radiation is applied on vocal cords with a micromanipulator (Acuspot 711 Sharplan) under microscopic control and the laser beam had a Gaussian profile. For the irradiation was made two sets of tissue samples, one for the OCT analysis and, one was sent to the histological laboratory for the comparative analysis. The two sets of samples were made in the same condition and were irradiated with the same laser parameters: power, pulse duration and beam shape. After the irradiation, one set of sample was analyses with the OCT (Spectral Radar Coherence Tomography -SROCT ThorLab version) and we obtained images with the ablation crater, i.e., width and depth of the laser ablation crater. The refractive index of porcine vocal cords, for the OCT measurements was n = 1.39 [15]. Each OCT image is an optical cross-section of the tissue sample, and was obtained with a time delay not larger than one hour from the irradiation process. Optimal histological analyze requires a procedure which can take few days. This can cause modifications in the sample structure. For histological analysis, the tissue was fixed in formaldehyde, serially dehydrated in graded ethanol baths, embedded in paraffin, and sectioned in micro sections with thickness of ~ 6 μm. Digital images were obtained with an optical microscope.

In the first part we present the experimental results, the crater ablation size obtained on the ENT soft tissue irradiated with a CO_2 laser, analyzed with OCT and histological. In the second part, we use the numerical analysis to study the temperature distribution in the soft tissue after laser irradiation.

In order to obtain the power density we have measured the focal spot from the OCT images. The value obtained using the software from the SROCT – Thor Lab was of 800 μm. The data were used in numerical simulation to compute the heat transport in one dimension, i.e., into the deep of the tissue. For numerical model we have used to describe the tissue thermal properties the values of the liver tissue (see Table 1) and we made the assumptions that the tissue is homogeneous and isotropic [16, 17].

Table 1: Parameters used for numerical simulation.

density ρ (kg/m^3)	specific heat C_p (J/kg*K)	thermal conductivity k (W/m*K)	initial temperature T_0 (K)
1060	3600	0.52	310.15

3. Results and discussion

In the OCT images, the high reflectivity at the burn surfaces is due to carbonization zone. OCT and histology showed no significant differences, the crater depth is directly proportional to laser power, the ablation crater has a conic shape, with an input aperture of about 800 µm diameter and a depth of ~ 1 mm, which is in accordance with the focal length of the laser beam focusing system (f = 125 mm) and with the laser beam divergence. The crater walls presents necrosis with a thickness of 60-80 µm (the necrosis thickness was evaluated taking into account the diameter (6-7 µm) of the hemoglobin cells and lymphocytes cells visible in the image of the optical microscope and in the OCT images we measure the crater ablation dimensions using the corresponding software.

Figures from left side shows the histological images obtained at different laser beam powers with the same exposed time i.e., t = 100 ms. As one can see the ablation volume and the thermal damage in the surrounding tissue increase with the power of the laser beam.

At the lower power of the laser beam i.e., the 6 W, the irradiation effects are observed only as a necrosis zone there is no ablation crater in the tissue (see Fig. 1a-b). By increasing with more than a factor two the power of the laser beam, i.e., 14 W, a smooth crater in a zone with muscle from the vocal cords is observed (see Fig. 1c - d). For this sample there is no zone of necrosis because it was past a long time from the moment of the vocal cords porcine extraction. At 16 W of laser beam power, one can see the crater ablation, with a small zone of necrosis on the edge, the lymphatic system is intact, and there is no damage caused by the heat propagation (see Fig. e –f. At high power, above 20 W, the histological images show a clear crater ablation with small necrosis zone and with visible hemoglobin and lymphocytes cells (see Fig.1g -h obtained for 20 W and Fig. 1i - j for 24 W).

Figures from right side shows the OCT image obtained at different power laser beam with the same exposed time i.e., t = 100 ms, i.e., the same experimental parameters used for the tissue irradiation with the laser beam, as those used in histological analysis. The irradiation area is indicated by a bleak ring on the figures. The image observed at small distance, below approximately 1.5 mm, is due to the junction of the vocal cords.

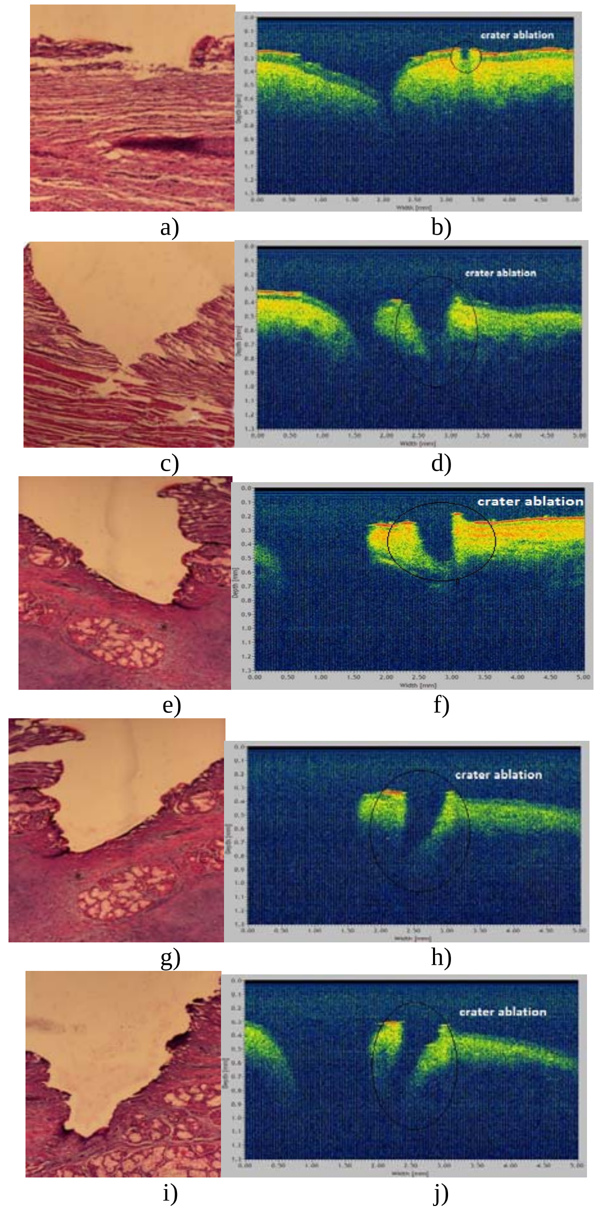

Fig. 1. Porcine vocal cords ablation by CO_2 laser: a) histological image P = 6 W and t = 100 ms, b) OCT image P = 6 W and t = 100 ms; c) histological image at P = 14 W and t = 100 ms; d) OCT image at P = 14 W and t = 100 ms; e) histological image at P = 16 W and t = 100 ms; f) OCT image at P = 16 W and t = 100 ms; g) histological image at P = 20 W and t = 100 ms; h) Oct image at P = 20 W and t = 100 ms; i) histological image at P = 24 W and t = 100 ms; j) OCT image at P = 24 W and t = 100 ms

As one can observe from figures 1, there is a very good agreement between the histological and OCT images. The effects, e.g., the necrosis zone and the ablation crater, and the proportionality between the laser power and depth of the crater are obtained [18].

In order to simulate the laser beam interaction with the soft tissue, we have used in our numerical simulation the same parameters as those used in experiment set-up and the expose time used was 100 ms, the spot laser beam 800 μm and the power density calculated for each laser power. To analyze the laser beam effects on the tissue, we have used a temperature value of 100 ^{0}C as the temperature at which the laser ablation begins [1, 19, 20].

Fig. 2 a – e shows the results obtained from numerical analysis for temperature distribution along the laser beam propagation. By increasing the laser power from 6 to 20 W the surface temperature, i.e., the maximum temperature increases from 67 ^{0}C to 138 ^{0}C. As was observed from the histological and confirmed by OCT analysis, at 6 W laser beam power, no ablation crater can be observed, the maximum temperature reached by the tissue is too low (see Fig. 2.a). At 14 W power laser beams, the surface temperature value is 108 ^{0}C, and it is obtained a temperature above 100 ^{0}C for approximately 14 μm along the laser beam. Increasing the laser beam power to 20 W, the distance which presents a temperature above 100 ^{0}C is 56 μm and 74 μm for 24 W laser powers. This results are confirmed by the histological and OCT measurements, which indeed evidenced a clearly ablation crater above 20 W.

Fig. 3 shows the crater ablation depth, i.e. the distance with the temperature above 100 ^{0}C along the laser beam propagation at different power and at an exposed time of 100 ms.

At this short exposed time, a linear dependence of the ablation crater deep from the laser power is observed. From the best fit parameter we have obtained a grow rate of 6.2 +/- 0.4 μm/ W. Moreover, we have obtained the beginning of the ablation crater, i.e., a temperature above or equal to 100 ^{0}C for a laser power of 11.3 W. With this optical parameter, thermal properties of the soft tissue and exposed irradiation time, there are no ablation below 11.3 W. Those results are in a good agreement with the values observed by histological and OCT measurements.

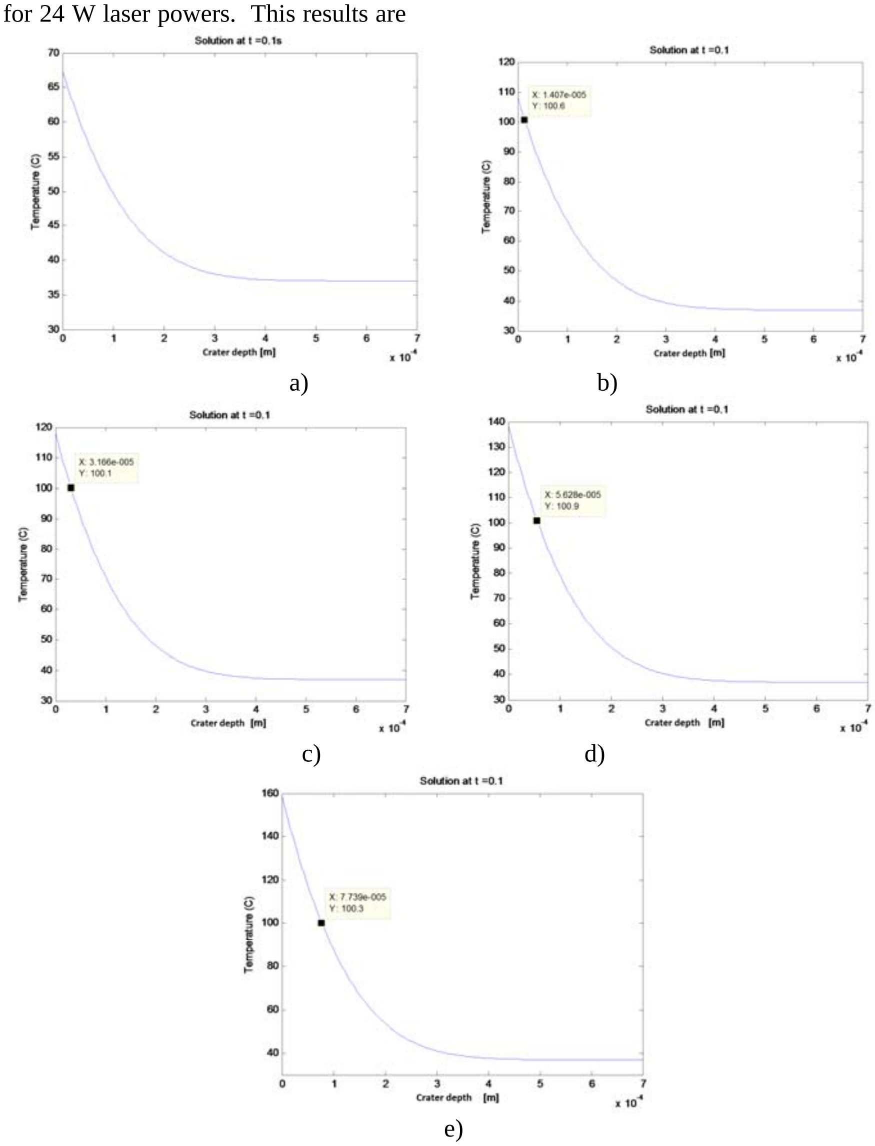

Fig. 2. Temperature distribution in the crater ablation depth and the maximum temperature at t=0.1 sec: a) P= 6 W, b) P=14 W, c) P=16 W, d) P=20 W, e) P=24 W

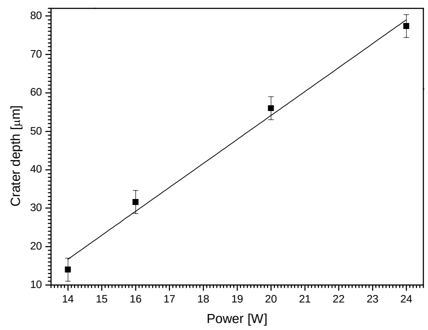

Fig. 3. The ablation crater depth at different laser beam power. The distance for which the temperature is above $100\,^{0}C$.

4. Conclusions

Comparison between the optical coherence tomography and conventional histological analysis demonstrate a very good agreement in the evaluation of the CO_2 laser injury of porcine larynges. The grate advantages offered by the optical coherence tomography can be used in the monitoring of the CO_2 laser surgery procedures.

A very simple model was used to describe de temperature distribution in the soft tissue. The numerical results obtained are in good agreement with the optical coherence tomography and conventional histological experimental data. Moreover, given the experimental parameter the threshold value for the ablation process of the CO_2 laser beam power was found.

The OCT can be an ideal method for real-time monitoring of vocal fold microsurgery due the adequate imaging depth and its high resolution and the surgeon could realize the therapeutic protocol using the simulation methods for the wanted result.

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*Corresponding author: mioara.petrus@inflpr.ro