Prediction of Automated Evaporation of Soft Biotissues of Different Types by Continuous CO_2 Laser Radiation

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Abstract. The study of laser evaporation of biological tissues and the creation of computational models of this process is an important scientific task for the development of predictive modeling and preoperative planning approaches in robotic laser surgery. Laser evaporation of soft water-containing biotissues with significantly different structure and strength characteristics was studied in the process of two-coordinate laser scanning of continuous CO_2 laser radiation. Samples of soft tissues from pigs *in vitro* (myocardium, liver, skeletal muscle, aorta) were used. It is shown that the dependence of the laser evaporation depth on the radiation power for these tissues is well approximated by a linear function. The proportionality coefficients between the maximum evaporation depth and the radiation power for all tissues, with the exception of the aorta, practically coincide with the calculated coefficient of 175 μm/W, obtained on the basis of the evaporation model of soft tissue evaporation for scanning parameters: scanning speed v = 15 mm/s, laser beam diameter 180 μm, distance between lines 150 μm. It is shown that despite the strong differences in soft tissue structure and strength characteristics (more than two orders of magnitude), the average deviation of the calculated value of the evaporation depth from the measured value for all tissues is several tens of percent and depends on the excess over the evaporation threshold. The developed approach can be used for predictive modeling in preoperative planning using scanning systems based on continuous CO_2 lasers.

Keywords: CO_2 laser; automated scanning; biotissue; structure; strength characteristics; depth of evaporation; predictive modeling.

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1 Introduction

Robotics in surgery is one of the directions of development of modern medicine [1–3]. It is assumed that robotic surgery will reduce the risks of the human factor and shorten the time of operations, increase the precision of surgical intervention and the effectiveness of treatment, and accelerate postoperative rehabilitation of patients.

One of the tools widely used for high-precision and minimally invasive laser surgeries are CO_2 laser-based devices equipped with scanning systems. Such devices allow for the automation of the laser radiation exposure process [4]. The most common use of such devices is for the so-called ablative (local tissue evaporation) and non-ablative (local tissue heating) laser skin resurfacing [5,6]. Such technology is considered to be the gold standard in dermatological surgery. Currently, active research is being conducted on the use of scanning laser systems in relation to the tasks of obtaining precision cuts of complex shape of a given depth and evaporation of a given area of arbitrary shape on the surface of biotissue. Scanning systems allow precise control of laser radiation parameters in the impact zone. They ensure high speed and accuracy, safety and reproducibility of operations using CO_2 lasers [7]. This significantly expands the possibilities and prospects for their use in surgery.

One of the directions of development of robotic laser surgery is predictive modeling for preoperative planning purposes [8, 9]. This approach is based on experimental and numerical modeling of laser exposure processes using various *in vivo* or *ex vivo* biomodels. In Ref. [10], the selection of modes and prediction of the depth of automated laser incision of soft tissues is proposed to be

carried out on the basis of an experimental database. To create such a database, chicken muscle tissue *ex vivo* was used. In Refs. [11, 12], to obtain a given depth of laser evaporation of soft and bone tissues, it is proposed to use simplified mathematical models for predicting the depth of the cut based on experimental data on animal tissues. In Ref. [13], a 3D model based on a single-layer perceptron network is proposed to predict the creation of a cavity in tissue during laser ablation. Thus, to solve problems of prediction and optimizing laser ablation of biotissues, there is a tendency to use approaches based on experimental databases. This is due to the rapid development of information technologies and various data processing methods, including neural network-based methods. Another reason is the difficulty of direct physical modeling of laser ablation of biotissues. This is due to the fact that various interrelated processes occur in the laser exposure zone: absorption/scattering of radiation, thermal conductivity, chemical processes, explosive boiling, generation of acoustic and shock waves, etc.

Approaches to modeling physical processes during laser ablation of biotissues have been developing over several decades. In Refs. [14, 15], an analysis of laser ablation of soft biotissues by pulsed radiation for different wavelengths is carried out. For assessments, a thermal model is used and the overheating of water in soft tissues is taken into account. In Ref. [15], calculation expressions are given for the threshold intensity of ablation of biotissues and the depth of ablation by a short laser pulse. A logarithmic dependence of the ablation depth on the radiation intensity was obtained. In this work, the literature data on the depth of pulsed ablation of biotissues for different wavelengths are analyzed and compared with the proposed calculation formula. In some cases, good agreement between the experimental data and the calculated dependence was observed. In Ref. [16], the influence of the strength characteristics of soft biological tissues on the efficiency of their ablation by a single CO_2 laser pulse was investigated. *In vitro* tissues of animal myocardium, aorta, liver and skin were used. The water content in these tissues was approximately the same (70−80%), but the tensile strength differed by one to two orders of magnitude (the lowest in the liver, the highest in the skin). A direct relationship has been established between the efficiency of pulsed ablation (mass removal per pulse) and the strength characteristics of these tissues. It has been shown that under conditions significantly exceeding the threshold, the greatest ablation efficiency is in the liver, 3 times less in the myocardium, and 7 times less in the aorta. It is concluded that not only water but also strength properties play an important role in soft tissue ablation with a pulsed CO_2 laser. In Ref. [17], a significant influence of strength properties on the threshold of pulsed laser ablation of soft tissues was theoretically predicted.

It should be noted that the majority of studies are devoted to laser ablation of biotissues in short pulse mode. Such impacts create excess pressure, leading to tissue rupture. In this regard, the strength properties of tissues play an important role in the process of their laser ablation with short pulses. There are far fewer studies that examine the effects of continuous laser radiation on biotissue. Moreover, there are few works that model the cutting depth and the depth of evaporation of the tissue zone during two-dimensional scanning of continuous laser radiation. There is a well-known theoretical work [18], which presents a model of laser ablation of biotissues for continuous radiation with a stationary position of the laser beam. The model is based on thermal conductivity and the energy required for water evaporation. Based on the heat conductivity equation, relationships were obtained for the ablation threshold (when the temperature is equal to the boiling point of water for an exposure time of ∞) for the radiation intensity and exposure time required to begin ablation.

Previously, we proposed a semi-empirical model of laser evaporation of soft biotissues using continuous CO_2 laser radiation during two-coordinate beam scanning [19]. This model showed good agreement with experimental data on the evaporation of model samples based on aqueous solution of agar and samples of pig heart muscle *in vitro*. The model is based on energy balance, assuming that the key factor in energy balance is tissue water evaporation. The paper suggests that the proposed model can be used to predict the evaporation of other soft tissues with high water content during two-dimensional scanning of CO_2 laser radiation.

However, the applicability of the evaporative mechanism of tissue removal by continuous CO_2 laser for different soft tissues has not been definitively substantiated. This is due to the fact that soft tissues, despite similar thermophysical characteristics, have different structures. The different composition of the organic matrix and its structure leads to the fact that the strength properties of soft tissues differ significantly. For example, the tensile strength of soft tissues can differ by more than two orders of magnitude [16, 20, 21], while the maximum difference in thermophysical characteristics is tens of percent. In this regard, the question arises about the applicability of the evaporation model for soft tissues with very different strength characteristics under continuous laser radiation.

The aim of this work is to study the depth of evaporation of soft water-containing biotissues with very different strength characteristics during two-coordinate laser scanning of continuous CO_2 laser radiation, to predict the depth of evaporation based on the evaporation mechanism and to determine the limits of applicability of this approach.

2 Methods and Materials

For automated surface laser evaporation of biomodels, the setup was used, consisting of a laser unit, an articulated-mirror manipulator and a 2- coordinate galvanic scanner JS1105 Galvo Scanner (Sino-Galvo Technology Co., China) coupled to it. The setup is based on a single-mode CO_2 laser C-20A (Coherent, USA) with a power of up to 20 W with high-frequency (HF) pumping of the active medium. A detailed description of the setup is given in Ref. [22].

Table 1 Some characteristics of biological tissues [16, 20, 21, 23−25].

	Thermophysical			Mechanical and structural	
Type of tissue	**Thermal conductivity k, W/m·K**	**Heat capacity C, J/kg·K**	**Density ρ, g/cm^3**	**Tensile strength σ_p, N/mm^2**	**Peculiarities of the structure of biotissue**
Skeletal muscle	0.45	3450	1.04	0.07 (along the fibers)	An aggregate of parallel muscle fibers in connective tissue.
Myocardium	0.71	3730	1.06	0.04	Three-dimensional structure of densely packed muscle fibers.
Liver	0.52	3600	1.06	< 0.04	An aggregate of liver lobules and hepatocytes.
Aorta	0.63	3960	1.04	2	A set of elastic functional shell layers.

The scanner control unit and our specially developed software allowed us to set any shape of the evaporation area within 20 mm × 20 mm, the speed of the laser beam within 1–50 mm/s, and the required filling density of the evaporated surface by laser hatching. The surface scanning area is a set of parallel lines (hereinafter, hatching lines). The required density of filling the evaporated area of the surface by laser hatching was ensured by changing the distance between the hatching lines.

The following soft tissue samples of pigs *in vitro* of different types with high water content (70−80%) were used as biomodels: cardiac muscle (myocardium), liver, skeletal muscle 4−5 mm thick and aorta 3−3.5 mm thick. These biotissues are similar in water content and thermophysical characteristics, but differ significantly in structure and strength characteristics (Table 1). The biotissues were used for 48 h post mortem. Between experiments, all samples were stored in a closed Petri dish at a temperature of 0−4 °C.

In experiments on automated evaporation of biomodel surface areas, a single scan of a rectangular area measuring 8.5 mm × 11.6 mm was performed. The distance between the hatching lines was 150 μm. A lens with a focal length of 50 mm was used. The radiation was focused on the surface of the biomodel. The focal spot size was 180 μm at a level of 86%. In the experiments, the evaporated zone of biotissues was studied depending on the power of laser radiation at a constant scanning speed of 15 mm/s.

After scanning, the samples were frozen in a refrigerator at −15 °C. Then, transverse sections of the scanning area were obtained in these samples. The cut was made with a blade through the center of the evaporated zone perpendicular to the surface of the sample and in a direction perpendicular to the direction of the laser scanning lines. The evaporation depth of the samples was measured using photographs of cross-sections of the evaporation zone. The evaporation depth was determined for 5 samples at fixed values of power, scanning speed and hatching density. Then the obtained values were averaged and the standard deviation was found.

3 Results

Fig. 1 shows cross-sections of the evaporated zone of different types of biological tissues at a radiation power of 6.5 W and a scanning speed of 15 mm/s. It can be seen that the section of the evaporated zone has a sawtooth shape. This is due to the incomplete overlap of individual cuts when scanning the laser beam. Also, additional irregular unevenness of the evaporated zone is caused by the heterogeneity of biotissues and the peculiarities of non-stationary processes of mass removal from the zone of laser radiation exposure [26, 27]. In Fig. 1, it can be seen that for skeletal muscle, the shape of the “teeth” is irregular, some of the “teeth” stick together, and their surface is loose. A similar picture was observed for the heart muscle and liver. For the aorta, the shape of the evaporated zone has clearer outlines, and the “teeth” are more regularly distributed across the evaporated zone. It is obvious that such a qualitative difference in surfaces is caused by the fact that the aorta is a much stronger tissue compared to others. Under the saw-tooth surface of the evaporated tissue zone, a zone with a whitish tint was observed, which is associated with tissue coagulation. This zone was not observed for the aortic tissue, since the original aortic tissue already has a light whitish color.

As in work [19], the result of evaporation of biotissues during laser scanning was characterized by the maximum depth H_{max}. The characterization of the scanning result by maximum depth is caused by the difficulty of determining the average evaporation depth $\langle H \rangle$ from photographs of sections. The H_{max} value was averaged over sections of 5 samples under the same scanning conditions. Fig. 2 shows the measured depths of evaporation of skeletal muscle and aorta depending on the power of laser radiation. As can be seen, these dependencies are linear. A similar pattern was observed for the other biological tissues. Moreover, the maximum evaporation depth was obtained for skeletal muscle, and the minimum evaporation depth was obtained for the aorta. On average, the evaporation depth for skeletal muscle was 42% greater than for aorta. For other tissues, the difference in evaporation depth was smaller.

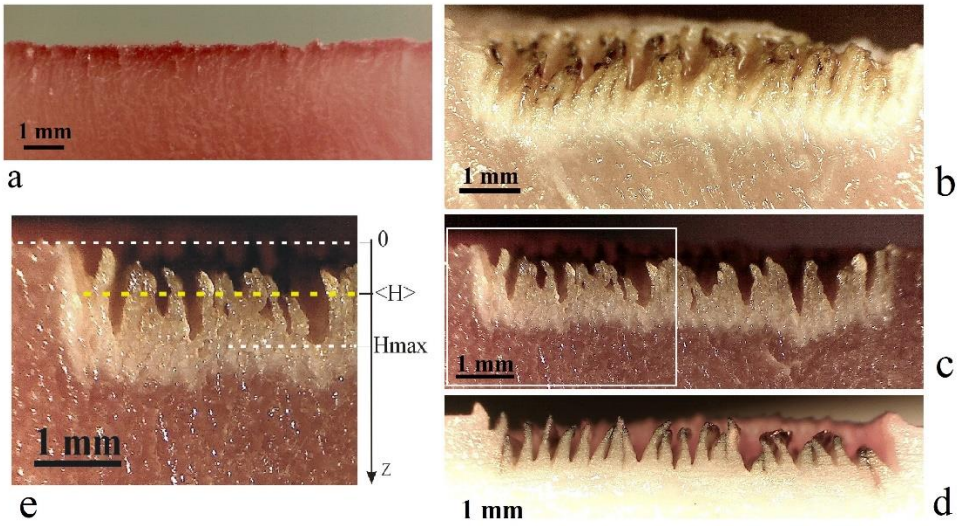
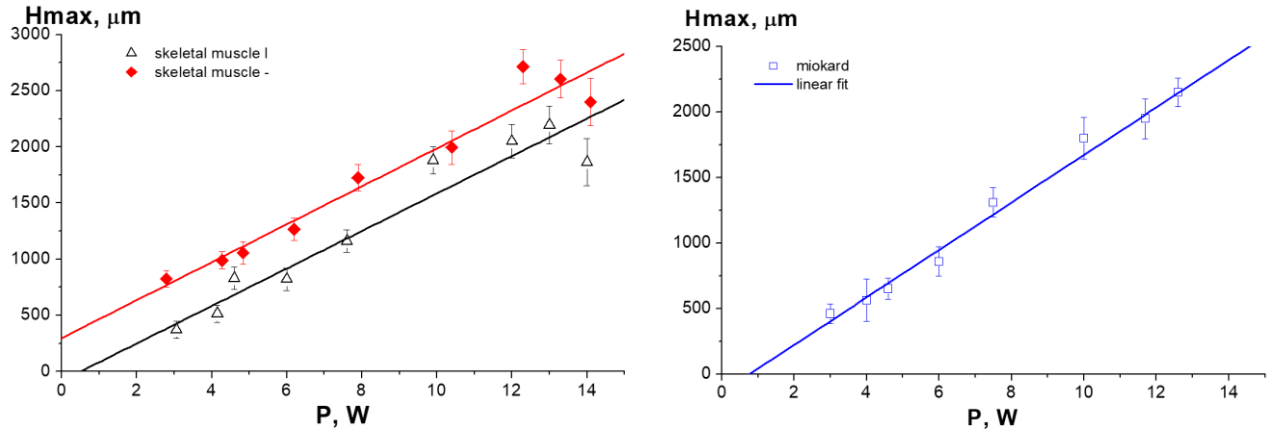


Fig. 1 Photographs of cross-sections of biological tissues before and after laser exposure at a radiation power of 6.5 W and a scanning speed of 15 mm/s. (a) Cross-section of skeletal muscle before laser irradiation; (b−e) cross-sections after laser irradiation; (b) skeletal muscle, muscle fibers were located vertically; (c) skeletal muscle, muscle fibers are directed horizontally along the scanning line of the laser beam; (d) aorta; (e) enlarged scale for the evaporation zone of skeletal muscle, which is indicated by the frame in panel (c). <*H*> − average evaporation depth.



(a) (b)

Fig. 2 Dependence of the maximum depth of evaporation of (a) skeletal muscle and (b) myocardium on the power of laser radiation at a scanning speed of 15 mm/s and a distance between strokes of 150 μm. White triangle marks – fibers along the laser beam (vertically (|)); red rhomb marks – fibers horizontally (-) and along the scanning line of the laser beam.

As follows from Fig. 2(a), there is a noticeable difference in the depth of evaporation of skeletal muscle depending on the location of muscle fibers relative to the scanning trajectory of the laser beam. When the fibers are located along (|) the laser beam, the evaporation depth is less than the depth when the fibers are located horizontally (-) (perpendicular to the laser beam). At low radiation power, H_{max} differs by more than 2 times for different fiber arrangements, whereas at higher power, this difference is 25%. This indicates that not only water evaporation but also tissue matrix rupture, which depends on the arrangement of the fibers, plays a role in tissue mass removal. The lower depth of evaporation with a vertical arrangement of fibers can be explained by the fact that under excess water vapor pressure, muscle fibers do not break, but split. After the radiation exposure stops, the unbroken fibers “stick together”. When the fibers are arranged horizontally, excess pressure causes them to rupture and be carried away from the impact zone along with water vapor. These mechanisms of laser evaporation of anisotropic tissue require more careful consideration using histological methods.

For each studied biotissue, the coefficients of the linear dependence $H_{max} = a + b \cdot P$ were found using the least squares method. The results are presented in Table 2. Almost all linear dependencies have a close coefficient b in the range of 164–181 μm/W. The exception is the aorta, for which b is noticeably smaller.

Table 2 Parameters of soft tissue evaporation by continuous CO_2 laser radiation.

№	Biotissue	*a*, μm	*b*, μm/W	dH_{max}, %	*Pc*, W
1	Myocardium	−129 ± 59	181 ± 6	0.2−19	2.1
2	Liver	109 ± 100	176 ± 13	0.2−32	2.1
3	Skeletal muscle (-)	297 ± 140	169 ± 15	2.7−68	6.5
4	Skeletal muscle (|)	−60 ± 170	164 ± 18	2.3−42	1.4
5	Aorta	83 ± 54	125 ± 6	5−30	1.1

The coefficient *a* differs significantly more for different tissues. In our opinion, this is due to the fact that the coefficient *a* is more closely related to the tissue evaporation threshold. In this paper we do not consider the issue of tissue evaporation threshold. It is obvious that near the threshold the dependence of the evaporation depth on the power will not be linear. In this mode, many processes play a role, such as thermal conductivity, near-surface heat outflow from the surface due to surface evaporation of water, diffusion of intra-tissue water, the occurrence of elastic deformations of the tissue in the laser exposure zone, etc. For this reason, the coefficient *a* varies significantly for different tissues. Far from the evaporation threshold, when intensive evaporation of tissue water occurs, its boiling, accompanied by micro-explosions, the water-steam phase transition plays a major role in the energy balance. This is evident in the fact that the proportionality coefficients b between the evaporation depth and power are quite close for different tissues.

In work [19] we developed a semi-empirical evaporation model for calculating the depth of evaporation of soft tissues when scanning with a continuous CO_2 laser beam. According to this model, when scanning a laser beam with a distance between strokes *d*, an evaporated zone is formed on the surface of the tissue, having a structure with maxima and minima with a period *d*. The shape of such an evaporated zone, assuming a Gaussian shape of a separate cut of tissue, is described by the formula:

$$H(x) = \frac{P}{\sqrt{\pi} V \rho Q r_0} \sum_{k=0}^{N} e^{-\left(\frac{x-kd}{r_0}\right)^2}, \qquad (1)$$

where Q = 2.26·10^6 J/kg – specific heat of water vaporization, ρ – water density, P – laser radiation power, V – scanning speed, x – coordinate along the x-axis, directed perpendicular to the direction of laser lines during scanning

It can be seen that the depth of evaporation is proportional to the power *P*. In this case, the proportionality coefficient is determined by the tissue scanning parameters *V* and *d*. In accordance with Eq. (1), we obtained (using the Mathcad program) that at *P* of 1 *W*, *V* of 15 mm/s and a hatching period *d* of 150 μm, the maximum evaporation depth is 175 μm. Thus, the proportionality coefficient b_0 between the maximum evaporation depth H_{max} and the laser radiation power is equaled to 175 μm/W. This is practically identical to the coefficient *b* obtained experimentally for the liver, myocardium and skeletal muscle (see Table 2). For the aorta, this coefficient is significantly less than the calculated value. This is due to the strength characteristics of this tissue. In this case, a significant portion of the invested energy goes not only to the water-steam phase transition, but also to the rupture of the more durable aortic matrix.

In Fig 3, all measured depth values for all studied biotissues are shown on one graph. This graph also shows the calculated linear dependence $H_{max} = b \cdot P$ is equaled to 175 μm/W · P.

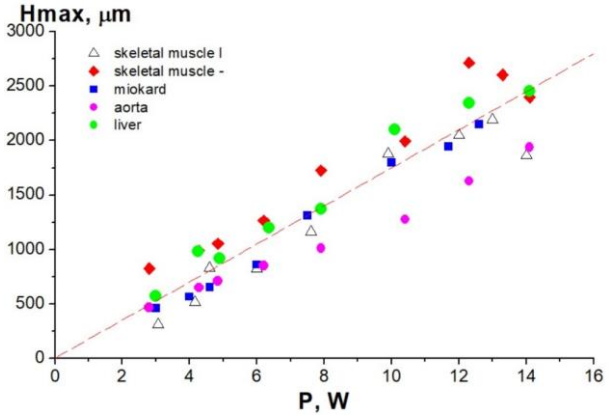


Fig. 3 Maximum depth of evaporation of soft biotissues. The dotted line is the calculated dependence according to the evaporation model of evaporation of soft biotissues as shown in Eq. (1).

Table 2 shows the range of variation in the deviation value dH_{max} of the measured depth H_{max} from the calculated linear dependence $b_0 \cdot P$ for different tissues. The maximum difference between the depth of evaporation of biological tissues and the calculated value is tens of percent. The standard deviation of the measured values of H_{max} for all biological tissues from the calculated linear dependence of H_{max} is 16%. Large deviations (e.g. 68% for skeletal muscle) between calculation and experiment occur in the low power region (up to 3−4 W). This is due to the fact that the calculated dependence does not take into account the evaporation threshold. In the evaporation model presented by us in Ref. [19], it is assumed that laser action occurs far from the evaporation threshold of biological tissue.

Let us estimate how the depth of evaporated biotissue differs from the calculated $H_{max}0 = b_0 \cdot P$ ($b_0 = 175$ μm/W) depending on the laser radiation power. Let for a specific biotissue $H_{max} = a + b \cdot P$. Then the relative deviation of this value from the calculated one will be:

$$\eta \equiv \left|\frac{H_{max}0 - H_{max}}{H_{max}0}\right| = \left|\frac{a}{b_0 * P} + \frac{b - b_0}{b_0}\right|. \quad (2)$$

The magnitude of this deviation is related to two components. The first is due to the fact that the tissue has a non-zero value of the parameter a. This first component decreases with increasing P, i.e. with distance from the threshold of the onset of evaporation. The second component of the deviation η is caused by the difference between the proportionality coefficient b and the calculated value b_0. As already noted, this coefficient for soft tissues is quite close and the value of $(b-b_0)/b_0$ is up to 12% (for skeletal muscle when the fibers are located along the laser beam) and 29% for the aorta.

From Eq. (2) we estimate the boundary power Pc, at which the deviation of the calculated dependence from the measured one will be 30%. Table 3 shows the Pc values for different tissues. For all tissues, the deviation of the evaporation depth dH_{max} from the calculated value is less than 30% when the radiation power Pc is more than 1−2 W. The exception is skeletal muscle, where the muscle fibers are directed perpendicular to the laser beam. In this case, the difference in depth from the calculated value of less than 30% is obtained at radiation powers greater than 6.5 W.

Thus, despite the fact that soft tissues can differ significantly in their strength characteristics (more than 2 orders of magnitude as in the case of the liver and aorta), the depth of evaporation by continuous CO_2 laser radiation far from the evaporation threshold is in good agreement with the evaporation model for soft tissues. The approach developed in Ref. [19] for calculating the depth of soft tissue evaporation during 2D scanning of a laser beam can serve as a good predictive model for preoperative planning using continuous CO_2 laser radiation.

4 Conclusion

The evaporation of soft water-containing biotissues with very different structure and strength characteristics under two-coordinate laser scanning of continuous CO_2 laser radiation has been studied. It is shown that the depth of evaporation of biological tissues is well described by a linear dependence on the laser radiation power. The proportionality coefficients between the maximum evaporation depth and the laser radiation power for all tissues, with the exception of the aorta, practically coincide with the calculated coefficient of 175 μm/W (for a scanning speed v of 15 mm/s and a laser beam diameter of 180 μm), obtained on the basis of the evaporation model of soft tissue evaporation. For the aorta, this coefficient is lower and is (125 ± 6) μm/W. This difference is due to the significantly higher tensile strength of the aorta compared to other soft tissues.

The data analysis showed that despite the significant differences in the structure and strength characteristics of soft tissues, the average deviation of the calculated evaporation depth from the measured value across all tissues is 16%. At the same time, at low laser radiation powers, the deviation of the evaporation depth from the calculated value for a specific tissue can be tens of percent. This is due to the fact that near the evaporation threshold, the dependence of depth on power is nonlinear, which is not taken into account in the evaporation model. It was shown that for all soft tissues studied, with the laser exposure parameters used in the experiment, the deviation of the evaporation depth from the calculated value is less than 30% for a radiation power of more than 1−2 W. The exception is skeletal muscle, where the muscle fibers are directed perpendicular to the laser beam. In this case, the difference in depth from the calculated value of less than 30% is obtained at radiation powers greater than 6.5 W.

Based on the conducted studies, it can be stated that significant differences (by an order of magnitude or more) in the strength characteristics of soft tissues do not lead to an equally strong difference in the depth of evaporation by continuous CO_2 laser radiation. The approach we present can be used to implement predictive modeling in preoperative planning using scanning systems based on continuous CO_2 lasers.

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Disclosures

All authors declare that there is no conflict of interest in this paper.

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