Model for indirect laser surgery

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Abstract: We present a theoretical model for laser cutting of biological tissue by a strongly heated fiber tip with a highly absorbing coating. A significant dependence of the cutting speed and cutting depth on the inclination angle of the scalpel to the surface when scattering exceeds absorption in the biological tissue is shown. Experimental evidences of this effect are presented. In the experiment, we used silica fiber with coating made of carbon and silicon organic varnish, the 0.97-µm wavelength laser and porcine skin. The additional opportunity to increase the efficiency of cutting by deposition of the absorbing layer on the tissue surface is considered.

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1. Introduction

In recent years, the progress in laser surgery has been mainly associated with the approbation of newly created laser sources [1–3] and the use of novel focusing elements [4] or coatings [5–7]. The effect of the laser radiation on a biological tissue has been theoretically considered in Refs [8–10].

The development of new methods of contact laser surgery on a soft tissue requires optimization of the very procedure of tissue incision. Investigation of the specificities of contact cutting with novel coatings changing the temperature distribution at the incision site, ‘indirect laser surgery’ [11], should be carried out. The coating most commonly used in practice [12,13] is the strongly absorbing coating applied on tip of the fiber, which leads to high absorption of light and strong heating of the silica scalpel. In this case, not only absorption of light in the medium, but also the thermal diffusion from the strongly heated silica fiber should be considered.

In what follows, we discuss a laser resection technique with contact cutting by a multimode fused silica optical fiber [14]. In that technique, biological tissue heated by laser radiation softens and, under the mechanical action of the silica core of the optical fiber,
moves apart so that a cut is formed. That is the principal difference of the laser resection from the laser ablation technique which considers biotissue elimination.

Further in our study we will deal with a continuous lasing mode, as the simplest one, and a wavelength of 0.97 µm from a surgical diode laser, at which scattering significantly exceeds absorption in biological tissue. The currently used light-absorbing layers applied onto optical fiber are carbon-based [15,16] and have an absorption of tens and hundreds of reciprocal centimeters.

During most presentations, a surgeon holds a fiber scalpel like an ordinary scalpel at an angle of 30°-50° to the incision site, especially in the contact mode. This angle is convenient when the absorption in the medium is high, e.g., when a CO₂ laser is used. When the size of the absorption area is greater or approximately equal to the size of the cutting part of the laser scalpel, one needs to consider the geometry of incision, as well as absorption and scattering at a given wavelength.

We will discuss contact laser resection by a silica fiber tip with a highly light-absorbing layer applied on this tip. A number of specificities should be taken into account here. First, the heating source is introduced directly into the tissue, i.e., the action comes from inside of the tissue (see Fig. 1). Besides, at the considered wavelength, scattering substantially exceeds absorption. Thus, the absorption of the scattered light should also be taken into account. This regime is observed for common IR diode lasers operating in the “therapeutic transparency window” from 0.7 µm to 1.8 µm [17].

![Fig. 1. Schematic representation of contact laser resection of biological tissue by a silica multimode fiber for a) obtuse angle α>π/2 and b) acute angle α<π/2. 1 – silica fiber, 2 – biological tissue, SAC - strongly absorbing coating. Arrow shows the direction of fiber movement.](image)

A key supposition here is that, according to Ref [18], the cutting is possible when tissue temperature reaches a certain value. The mechanism of resection is assumed to be as follows: first, preheating to a certain, relatively low (compared to the ablation, for example) temperature, and then mechanical separation of the heated softened tissue by a moving solid tip of the optical fiber. The exact value of this temperature for various tissues is unknown, but we assume that this value is in the range from the denaturation temperature of structural proteins of biological tissues (60°-70° C) [19] to the vaporization temperature of water present in the biological tissue (100° C).

Consequently, the logical assumption is that resection is faster if the scalpel moves to the preheated region than in the opposite direction. Thus, contrary to the established practice, the angle of inclination of the optical fiber during the resection should possibly be obtuse. The effect of the laser light incident at different angles on the surface of the media [20–22], including biological tissues [23], has been discussed. But in our case of resection the optical scheme is different, and a new consideration of the thermophysical problem should be carried out.
The aim of this paper is to construct a simple theoretical model describing the main characteristics of this process, to study the effect of the inclination angle of the fiber tip on the parameters of tissue resection, and to prove the results experimentally.

2. Theoretical consideration

Let us construct a simple model of contact laser resection of a biological medium allowing for estimating the effect of the inclination angle of the fiber tip on the cutting parameters. For simplicity, the following two-dimensional model will be addressed. Consider a semi-infinite homogeneous medium with absorption coefficient $\alpha_a$ and scattering coefficient $k_{scat}$, into which an optical fiber tip with diameter $d$ penetrates at a depth $h$ at an angle $\alpha$. Let the medium be highly scattering, i.e., $k_{scat} \gg \alpha_a$. In the subsequent calculations we assume the following characteristic values for the medium: scattering coefficient $k_{scat} = 100 \text{ cm}^{-1}$ and absorption coefficient $\alpha_a = 1 \text{ cm}^{-1}$, which corresponds to the known average values for human skin [24–26]. A thin layer of strongly absorbing coating (SAC) applied onto the fiber tip is assumed to be infinitely thin and heated due to absorption of part $A_{ SAC}$ of passing radiation. The remaining part of the radiation passes through the medium where it gets scattered and absorbed. The divergence angle of the passing beam is $\alpha_{div}$.

The medium is heated as a result of the thermal transfer from the SAC heated by laser light, as well as due to the absorption within the medium of laser radiation coming through the SAC. We suppose that at a certain temperature $T_{cr}$ some changes occur in the medium (softening), allowing subsequent smooth cutting of the region by the silica fiber tip. We assume that when the temperature of the medium exceeds the value $T_{cr}$ along the entire penetration depth of the tip, the tip can freely move in this direction, making a cut. Let us define the maximum speed of movement $V_{max}$ of the fiber tip in the medium at which the temperature in the medium in the direction of the tip’s movement exceeds $T_{cr}$ along the entire penetration depth of the fiber tip. Here we take into account that in the previously irradiated area of the performed cut the medium is missing at the penetration depth of the tip $h$.

It is obvious that for a given cutting depth, determined by the value of $h$, the movement with speed $V_{max}$ provides most efficient cutting conditions. Indeed, lower speeds lead to additional heating of the medium and can cause undesirable thermal damage, for example, can increase the cutting depth. On the other hand, too great a speed may not allow sufficient heating for mechanical cutting, which may lead, for example, to a decrease of the cutting depth due to mechanical action of tissue on the optical fiber. Too great a resistance of insufficiently heated medium may also lead to an uneven movement of the optical fiber along the medium, causing inhomogeneity of the cutting parameters.

In order to calculate the temperature distribution in the medium, we solve the heat diffusion equation [27] in time $t$ and Cartesian coordinates $x, y$ variables assuming the thermal conductivity coefficient of the medium $\kappa$ is constant. Within a reference frame fixed with the fiber tip it reads

$$c_p \rho \left( \frac{\partial T}{\partial t} - \nabla T \right) = \kappa \Delta T + \alpha_a I,$$  \hspace{1cm} (1)

where $c_p$ is the specific heat of the medium at a constant pressure, $\rho$ is the density of the medium, $T$ is temperature increment above room temperature, $\nabla$ is the speed of fiber movement relative to the medium (see Fig. 1), and $I$ is the radiation intensity in the medium.

To determine the temperature distribution in fused silica, we use an analogous to Eq. (1) heat diffusion equation with typical for silica parameters of specific heat $c_{pq}$, density $\rho_q$, and thermal conductivity coefficient $\kappa_q$. The light absorption in silica is neglected, movement is absent.

At the medium-air interface heat transfer will be neglected. At the side boundary with silica the condition of equality of heat flows is valid:
where $r$ is a coordinate perpendicular to the surface. Approximating SAC by an infinitely thin layer yields the following condition at the side boundary of the silica:

$$
\kappa \frac{\partial T}{\partial r}_{\text{medium}} = \kappa_q \frac{\partial T}{\partial r}_{\text{quartz}} + A_{SAC} f_i,
$$

where $f_i$ is the initial intensity of radiation in the optical fiber.

Since the medium we consider here is a highly scattering one, it is necessary to take into account the scattered component of radiation that comes from the optical fiber to determine $I$ in Eq. (1). The propagation of the scattered component in the highly scattering medium will be described in terms of the diffusion model [28,29]:

$$
\frac{\partial q}{\partial t} - \nabla q = D \Delta q - \alpha_c cq + k_{scat} f,
$$

(2)

where $q$ is the power density of the scattered component, $D$ is the diffusion coefficient of light in the medium, $c$ is the velocity of light in the medium, and $f$ is the intensity of radiation that emerges from the fiber and is the source of the scattered component. The total radiation intensity in the medium is $I = f + cq$.

The diffusion coefficient $D$ in the highly scattering medium can be written as $D = c/(3k_{scat}(1-\mu))$, where $\mu$ is the average cosine of the scattering. The drift term is negligible in the considered region of the fiber movement speeds. In the case of a divergent beam, we simplistically assume

$$
f = f_0 \exp\left(-\left(\alpha_c + k_{scat}\right)r\right) / \left(1 - 2r \tan(\alpha_{div})\right).
$$

The boundary conditions take into account the effect of reflection of light at the boundaries of the active medium:

$$
D \left| \frac{\partial q}{\partial r} \right| = \alpha_q q,
$$

where $|\partial q/\partial r|$ is the derivative module with respect to the direction perpendicular to the surface of the silica fiber. The sign is chosen appropriately. According to the paper [30] devoted to the theory of light propagation in disordered media, the coefficients in the boundary conditions can be estimated as follows:

$$
\frac{D}{\alpha_q} = \frac{2}{3k_{scat}} \frac{1+R}{1-R},
$$

where $R$ is a coefficient determined by the refractive indices of the scattering medium $n_m$ and of the external medium, i.e., silica or air.

The maximum speed value $V_{\text{max}}$ of the fiber for given values of the inclination angle $\alpha$ and cutting depth $h$ is determined as follows. The stationary temperature distribution in the medium is found for a selected speed $V$ in a selected geometry. Then for each value of $y$ over the entire cutting depth $-h \leq y \leq 0$ we find the maximum temperature $T_{\text{max}}(y)$ along the $x$ axis. According to the condition of the possibility of the cut, for each $y$ from the range being considered the maximum temperature should be no less than $T_{cr}$. Accordingly, if the minimum value $T_{\text{max}}(y)$ in the range $-h \leq y \leq 0$ is less than $T_{cr}$, the execution of the cutting is considered to be impossible, and therefore too high a speed has been chosen: $V > V_{\text{max}}$. On the
contrary, if the heating of the medium is too high, too low a speed has been chosen. Obviously, \( V = V_{\text{max}} \) when the minimum value \( T_{\text{max}}(y) \) with respect to \( y \) in the range \(-h \leq y \leq 0\) coincides with \( T_{cr} \).

The temperature distribution in the contact region is shown in Fig. 2 at the maximum speed \( V_{\text{max}} \). It is seen that the maximal temperature is reached in the tissue not at the cutting side of the fiber.

![Image](image_url)  
Fig. 2. The temperature distribution in the contact region at the maximum speed \( V_{\text{max}} \) for the parameters: \( \alpha = 0.2, h = 250\mu m \) 1 – silica fiber, 2 – tissue.

Figure 3 shows the dependences \( V_{\text{max}}/V_0 \), where \( V_0 = \kappa/(c_p \rho d) \), on the inclination angle \( \alpha \) of the fiber tip at cutting depths 250 \( \mu \)m and 400 \( \mu \)m for the following values of parameters: \( d = 400 \mu m, A_{\text{SAC}} = 0.3, \alpha_{\text{div}} = 0.5, n_m = 1.4, n_q = 1.49, c_p \rho = 4.5 \text{ J/K/cm}^3, \kappa = 6.3 \times 10^{-3} \text{ W/K/cm} \).

![Image](image_url)  
Fig. 3. The dependence of maximum speed \( V_{\text{max}} \) on the fiber inclination angle \( \alpha \) for penetration depths \( h = 250\mu m \) (black line) and \( h = 400\mu m \) (red line). \( V_0 = 3.5 \times 10^{-2} \text{ cm/s} \).

As seen from the figure, the maximum speed of the fiber movement, at which the cut can be made, increases with the growing inclination angle up to values \( \alpha \sim 135^\circ \).

3. Materials and methods

Dependence of the tissue destruction characteristics on the inclination angle of the fiber scalpel relative to the surface was studied experimentally by applying a constant driving force onto the cutting part. The cutting part of the laser scalpel is the tip of a silica fiber glass core cleared from the cladding layer.
The biological sample was ex-vivo porcine skin. It was stretched to provide smooth surface and placed vertically. The fiber tip resided and moved in the same vertical plane perpendicular to the skin surface. The tip was pressed into the tissue at a depth of 0.5 mm. The angle between the tip and the surface was fixed. The fiber movement downwards was due to the constant gravity force of 0.44 N provided by the additional weight connected to the fiber. The inclination angles \( \alpha \) were 45°, 90°, 135° with respect to the skin surface (see Fig. 1).

The sources of laser radiation were moderate-power LS-0.97 IRE-Polus lasers operating at a wavelength of 0.97 µm (absorption coefficient in water 0.47 cm\(^{-1}\)). Silica fibers were used with a silica core diameter of 550 µm without coating and with an absorbing coating [31].

4. Experimental study of cut parameters in biological-like media depending on the inclination angle of contact fiber scalpel

To qualitatively check the theoretical considerations, we conducted experiments to measure the speed and depth of the cut at different angles of the fiber scalpel. The most difficult condition for theory validating is to obtain the steady state mode of operation. For this, a mode with a constant pressure force on the scalpel was chosen.

The measured cutting speed and depth averaged over 5 experiments with the constant pressure force using laser radiation at a wavelength of 0.97 µm are presented in Table 1. The deviation of the average cutting depth was about 25%.

| Angle       | depth | speed  |
|-------------|-------|--------|
| 45°         | 8 mm  | 0.07 mm/s |
| Perpendicular | 4.5 mm | 0.19 mm/s |
| 135°        | 4 mm  | 0.36 mm/s |

It is seen from the table that the cutting speed considerably depends on the inclination angle of the optical fiber to the direction of movement. The speed of fiber movement at an obtuse angle (135°) to the direction of movement is about twice as high as in the case when the fiber is oriented perpendicular to the direction of movement, and is significantly (5 times) higher than in the case when the angle is acute (45°). As the cutting depth measurements are not very precise we can declare the qualitative dependence only.

It should also be noted that the depth of cutting by an optical fiber oriented at an acute angle to the surface is significantly greater than in the two other cases. This fact may be indicative of excessive additional heating due to laser irradiation of previously cut material, which leads to additional thermal destruction.

The experimental results confirm the supposition made on the basis of the theoretical model about the dependence of the speed of contact cutting of biological media on the angle of inclination of the optical fiber. Based on this result, practical recommendations can be made to optimize the process of contact laser resection of biological tissues. Namely, the orientation of the optical fiber at an obtuse angle to the direction of fiber movement provides increased speed of resection, resulting in reduced damage to adjacent tissues.

Bearing in mind the above model of contact resection, the following way of increasing the cutting speed may be suggested. Let us assume that initially a coating is applied on the surface of biological medium along the intended cut line, and the coating provides absorption of radiation incident on it at the operating laser wavelength (see Fig. 4). In this case, if the cutting depth \( h \) is not too large, part of the scattered radiation will reach the surface and get absorbed in it, which leads to additional heating of the surface and, consequently, of adjacent tissue. This process can be expected to further improve heating of the medium in the direction of fiber movement, thus providing an additional increase in the cutting speed. The effect of
this improvement on $V_{\text{max}}$, obtained within the above-discussed theoretical model, is shown in Fig. 5.

![Fig. 4. Schematic representation of contact laser cutting of biological tissue by a silica multimode fiber with absorbing coating applied on the tissue: 1 – silica fiber, 2 – biological tissue, 3 – absorbing coating applied on tissue, 4 – cutting area.](image1)

![Fig. 5. Maximum speed $V_{\text{max}}/V_0$ versus fiber inclination angle $\alpha$ in case of “clear” surface (black line) and in case of light absorbing coating applied on the surface (red line). Cutting depth $h = 250 \mu m$. $V_0 = 3.5 \times 10^{-2} \text{ cm/s.}$](image2)

5. Conclusion

A simple theoretical model for contact laser cutting of biological tissue by a fiber tip with a highly light-absorbing layer applied on this tip has been developed. The model considers that the fiber tip is emerged into the biological tissue and inclined to its surface. The model allows to calculate the optimal speed of the laser cutting for given inclination angle and cutting depth. It is predicted that in the case of the obtuse angle, the cutting speed is higher than in the case of the acute angle.

It has been experimentally observed that the parameters of the laser cutting of tissues depend on the angle between the fiber and its movement direction along the surface. The statement that the cutting speed is higher for the obtuse angle than for the right one (and obviously than for the acute one) has been confirmed in porcine skin samples.

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