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ORIGINAL ARTICLE

Online estimation of laser incision depth for transoral microsurgery: approach and preliminary evaluation

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Abstract

Background The use of lasers in transoral surgery enables precise tissue incision with minimal adverse effects on surrounding structures. Nonetheless, the lack of haptic feedback during laser cutting impairs the surgeon's perception of the incision depth, potentially leading to undesired tissue damage.

Methods This paper presents a novel approach, based on statistical regression analysis, to estimate the laser incision depth in soft tissue. User trials were conducted in a laser surgery set-up, to verify the effectiveness of online estimation of incision depth in supporting precise tissue cutting.

Results The estimation accuracy was verified on *ex vivo* muscle tissue, revealing a root mean squared error (RMSE) of 0.1 mm for depths ranging up to 1.4 mm. Online estimation of depth has the potential to significantly improve the incision control of users.

Conclusions The proposed approach was successful in producing estimations of laser cutting depth in *ex vivo* muscle tissue. Further investigation is required to validate this approach on other types of tissue. Providing depth estimation during laser cutting allows users to perform more precise incisions.

Keywords transoral laser surgery; robotic surgery; laser incision modelling; biomedical robotics

Introduction

Laser technology is commonly employed in transoral surgery of the larynx (1). The purpose of these interventions is the treatment of malignancies that may occur in the upper aerodigestive tract, e.g. glottic cancer. Access to the surgical site is obtained through a laryngoscope inserted into the mouth of the patient (Figure 1). Diseased tissues are resected by incisions along their boundaries, using a CO_2 laser (1). The infrared beam emitted by these laser sources is strongly absorbed by the water present in the tissue, resulting in simultaneous cutting and coagulation (1). Clinicians determine the resection lines visually in order to ensure the complete removal of the tumour. At the same time, surgeons try to limit the extent of healthy tissue involved, to preserve as much

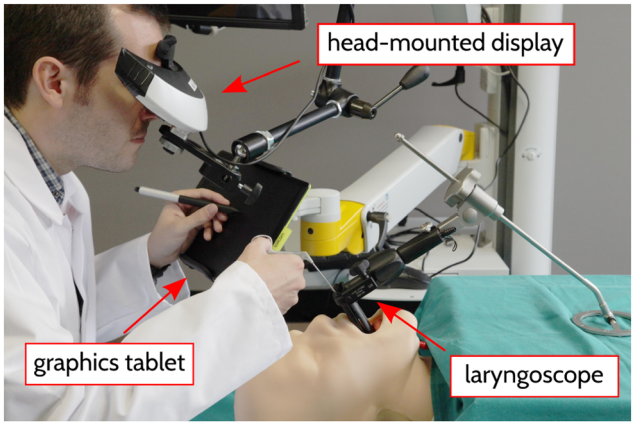

Figure 1. Mock set-up for transoral laser surgery of the larynx. Here, the surgeon uses a recently developed user interface (14), which allows him to visualize the surgical site through a head-mounted display (HMD) while controlling the position of the laser beam using a virtual scalpel (stylus and graphics tablet). A laryngoscope is inserted into the mouth of the patient. Diseased tissue is resected, firing the laser along an incision line. Forceps are used to apply traction to the tissue while the laser incisions are conducted

organ function as possible. The execution of accurate resections requires precise control of the laser cutting depth.

In current surgical equipment, incisions are controlled manually, moving the laser beam by means of a joystick and activating it through a footswitch (1). Current technologies available for transoral laser microsurgery (TLM) do not include any support for sensing the depth of laser cuts. The quality of incisions relies entirely on the dexterity and experience of clinicians. Extensive training is required to develop an effective laser cutting technique. This includes both the acquisition of basic knowledge of the physical principles behind laser ablation of tissue and the ability to manipulate the laser parameters and its exposure time in order to provide adequate cutting (1,2). Laser parameters used in clinical practice to regulate the incision depth include power, energy delivery mode, pulse duration and incision scanning frequency. The resulting depth of incision depends on the combination of these parameters plus the total time of laser exposure. However, no standard recipe exists to determine the laser parameters and the exposure time required to obtain an optimal incision. Physicians use different settings, depending on their skills, experience and preferred technique (1).

In this paper, we present a novel approach to implement online estimation of laser incision depth during cutting of soft tissues. The primary objective of this research is to enhance the perception of clinicians through the visualization of an estimated value of depth, thus supporting and enhancing their decision-making processes.

Different approaches have previously been proposed to estimate the amount of tissue removed during laser ablation. However, none of them fulfils the necessary requirements to offer online estimation of incision depth during TLM. Methods based on mathematical modelling (3) and numerical simulations (4) have provided a viable solution to predict the ablation volume in hard tissue. These methods assume a specific tissue composition and their applicability to soft tissue is yet to be verified. Furthermore, these approaches are computationally intensive and are thus not straightforward to implement in online applications.

Alternative solutions, based on sensing, have been developed recently (5,6). Such solutions provide the capability to track the ablation crater during laser irradiation, without any *a priori* knowledge of the physical properties of tissue; however, this approach requires the use of additional sensing devices in the proximity of the ablation zone. This limits their applicability in TLM due to space constraints, as the small size of the larynx does not offer enough volume for the introduction of additional equipment.

To overcome the limitations mentioned above, we propose to use a forward model, derived from the inputs used by the surgeons, to estimate the incision depth. In this paper we demonstrate that such a model can be obtained through statistical regression methods, using data captured during controlled laser experiments. To demonstrate the validity of this approach, we report on the generation of a model of incision depth in *ex vivo* muscle tissue. Experiments were conducted to explore the effects of different laser parameters on soft tissue and acquire data for the regression process. These experiments consisted of varying one parameter at a time and examining the resulting incision depth. Experimental data were used to generate the desired model, which served as the core of a prototype depth-estimation system. Here we present the experimental evaluation of this system, aimed at assessing its effectiveness in supporting precise laser incisions.

Materials and methods

We implemented a series of experiments under controlled conditions, using a surgical laser source to study the effects of the laser parameters on the resulting incision depth and to derive a model for online estimation.

Laser motion and activation were controlled by a computerized system and the resulting incisions were examined under a microscope to measure their depths. Two forms of target, gelatin phantom tissue and *ex vivo* chicken muscle tissue, were used in the study. A measurement protocol was implemented to obtain the input–output pairs required to derive the forward model, based on statistical regression techniques.

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Controlled incision of soft tissue

Incisions are produced by moving the laser beam along desired cutting paths on the tissue. Laser scanning, as described in (7), was used for this work: computer-controlled motorized mirrors are used to deflect the laser beam, enabling the automatic execution of preprogrammed cutting patterns. High-frequency cycles of the laser motion across the target tissue remove overlying layers with each pass.

When targeting laser light on biological tissue, different interactions may occur, e.g. thermal, photochemical, photoablation and photodisruption. The type of laser–tissue interaction depends on the power density (W/cm^2) and exposure time (s) (8). During laser cutting of soft tissue, a thermal interaction occurs: the laser radiation is absorbed by the tissue, resulting in a rise of temperature and the eventual evaporation of the water molecules in the tissue. Ablation by vaporization is the fundamental physical process governing laser incisions in soft tissue (8). The dynamics of this process can be controlled through the selection of appropriate laser parameters.

Different parameters of the laser source can be manipulated to influence the laser incision process. These include laser power (P) and scanning frequency (ω_s). In addition, the laser light can be delivered either as a continuous wave or through intermittent pulses. The influence of these parameters and that of the exposure time (t_{exp}) are studied by measuring the depths of incision obtained during controlled experiments. The results of these experiments are presented and discussed later in this paper.

The experimental set-up (Figure 2) uses a commercial surgical laser source, a Zeiss Opmilas CO_2 25 (wavelength 10.6 μm, TEM_{00} beam profile), whose power is configurable in the range 2–25 W. In the system used in this research, maximum power density is obtained when the laser spot is focused with a radius of 250 μm. Two energy delivery modes are provided, continuous wave (CW) and repeated pulse (RP), with the pulse durations (τ) 0.05, 0.1, 0.2 and 0.5 s. The concept of energy delivery mode will be further described in the next section. The CW/long-pulsed laser source used in this study has been superseded in clinical practice by short-pulsed (ms) lasers, which are known to produce more efficient cutting and reduced thermal damage (1). However, the use of this equipment does not limit the applicability of the methodology developed here. The estimation of incision depth proposed in this paper relies on a forward model that maps the laser inputs to the resulting incision depth. Thus, it can be applied to any laser source, provided that an appropriate forward model is used.

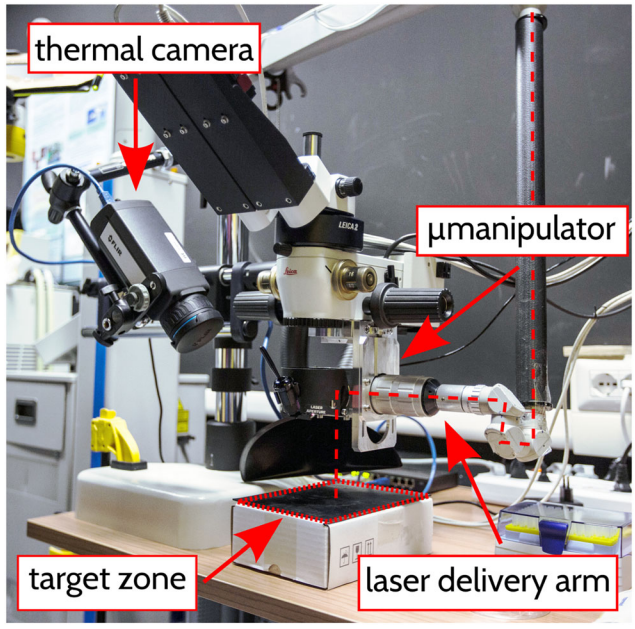

Figure 2. Experimental set-up. Tissue specimens were placed within the target zone. An infrared thermal camera (FLIR A655c) was used to monitor the temperature of tissue targets before the experiment. The laser beam was delivered on the target through a passive articulated arm. A motorized micromanipulator provided the means to control the laser motion

A motorized micromanipulator system previously developed in our laboratory (7) provides the controlled motion of the laser. This device regulates the motion of the laser through a tip/tilt fast-steering mirror (S-334, manufactured by PI GmbH, Germany), with a maximum accuracy of 4 μm at a distance of 400 mm (normal operating distance) from the target. The laser source and the micromanipulator are both connected to a Digital Operation Module (E-517, PI GmbH, Germany), which executes high-level commands related to the motion and activation of the laser. A user interface running on a GNU/Linux workstation allows the user to select exposure time, incision length and scanning speed.

Tissue targets

Initial incision trials were conducted on tissue phantoms. Cylindrical agar-based gel targets were produced to mimic soft tissue. The constituents used to fabricate these targets were deionized water and agar powder (B&V s.r.l., Italy); the concentrations were 98% water and 2% agar. In soft tissues, the absorption of infrared laser radiation is dominated by the presence of free water molecules (8), therefore water was chosen as the main constituent of the targets. Although different from tissue, these gels offer a controlled medium on which the effects of the laser can be reproducibly studied (9).

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In order to gather the data required to implement online estimation of laser cutting depth, additional incision trials were performed on fresh *ex vivo* chicken muscle tissue. Like most soft tissues, chicken breast has a high water content, which makes it a suitable target for CO_2 laser ablation trials. Before the experiments, tissue samples were kept for 20 min in an open refrigerated box at a controlled temperature (7–12 °C), in order to preserve their moisture and prevent degradation. To ensure identical initial temperatures, the samples were monitored with an infrared thermal camera (Figure 2). It is important to point out that the *ex vivo* model selected in this study does not present the same laser absorption profile that would be found in *in vivo* tissues, since the thermal effects of a laser on living tissues are influenced by factors that are not present in *ex vivo* models (8), e.g. heat convection due to blood perfusion. Nonetheless, most of these factors can be neglected at first approximation (8). The selection of an *ex vivo* model is consistent with the objective of this study, i.e. to prove the concept of online depth estimation based on statistical regression analysis.

Measurement of depth of incision

To examine the ablation craters, we used a digital microscope (Olympus SZX16). In order to obtain a complete exposure of the crater profile, the tissue targets were sectioned into slices.

Agar-based targets were sectioned manually with a blade. *Ex vivo* soft tissue targets required a more cautious protocol, as manipulation and sectioning may induce deformation artifacts that alter the measurements process. To preserve the structural properties of these samples throughout the examination, a snap-freeze technique (10) was used. This involves rapidly lowering the temperature of samples to −70 °C, by means of dry ice. This technique increases the rigidity of the specimens, allowing for sharp and clean slicing with minimum deformation. The targets were sectioned with a cryostat microtome, which performs the slicing while keeping the temperature of the samples low (−30 °C).

The depth of the incision is defined as the distance from the surface to the bottom of the incision crater (see Figure 3). This was measured by manual segmentation of the microscope images: the depth of the incision is estimated by contrasting its size in pixels against a reference scale bar.

Effects of laser parameters

In this section we describe the experiments conducted to determine the influence of the energy delivery mode and the frequency of the laser scanning motion on the depth of incision produced by the laser exposure.

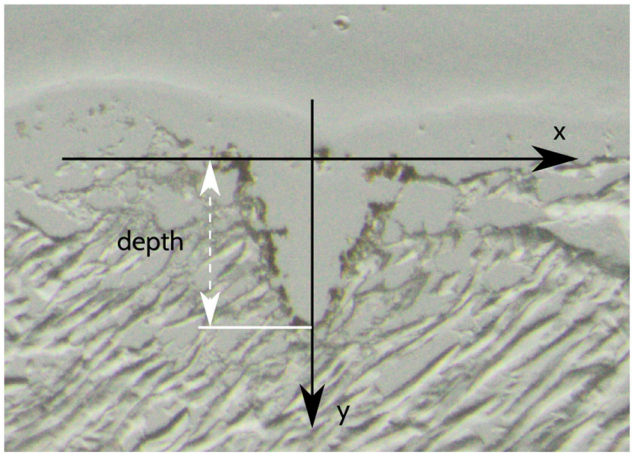

Figure 3. Transverse profile of an incision produced on *ex vivo* chicken tissue ($P = 3$ W, CW, $\omega_s = 10$ Hz, $t_{exp} = 3.5$ s). The origin of the reference frame is located at the middle of the incision on the surface of the sample

Influence of energy delivery mode

Prior research in the domain of laser–tissue interactions has established that the total amount of tissue removed is proportional to the energy delivered to the tissue (8). During TLM, the total energy transferred can be controlled by selecting a combination of laser power, delivery mode, pulse duration and exposure time.

A laser beam firing on the surface of a target produces a local irradiation, i.e. power density (W/mm^2), described by:

$$I = \frac{P}{A}$$

where P (W) is the emission power of the laser source and A (mm^2) is the area covered by the laser radiation. The amount of energy transferred to a target kept under irradiation for a time t_{exp} is defined as the energy density E (J/mm^2) (8).

Different delivery modes can be used to modulate the irradiation process. In general, a delivery mode is defined by a function $m(t)$, which affects the energy density being delivered to the target according to the following equation:

$$E = \int_0^{t_{exp}} I\, m\,(t) dt$$

An example is shown in Figure 4, where the effects of CW and RP delivery modes are contrasted. Using pulsed irradiation, the energy delivery process is slower, requiring additional exposure time to produce the same amount of energy density. When using the RP mode, the pulse duration (τ) needs to be configured. The laser source used in our experiments does not allow for an arbitrary value of duty cycle. Given a pulse duration, laser pulses are produced according to the following modulating function:

DOI: 10.1002/rcs

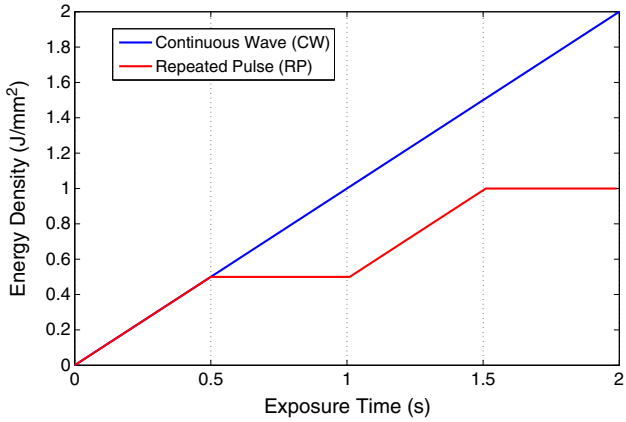

Figure 4. Power densities transferred to tissue by a laser source ($I = 1\,\mathrm{W/mm^2}$) by means of continuous wave (CW) and repeated pulse (RP) modes. For the RP mode, a pulse duration of 0.5 s and a duty cycle of 50% are considered

$$m(t) = \sum_k (-1)^k H(t - k\tau), k \in \mathrm{N}$$

where $H(t)$ is the Heaviside step function.

The quantification of the effects of delivering the same amount of energy through different delivery modes was performed experimentally. To exclude the effects of laser motion on the resultant depth, single point incisions were performed, i.e. no scanning motion was applied to the laser beam ($\omega_s = 0$). For these experiments, laser power was kept constant ($P = 2\,\mathrm{W}$).

We compared the effects of CW and RP ($\tau = 0.5\,\mathrm{s}$) for increasing values of energy. The experiment involved three pairs of experimental conditions. For each pair, we fixed the amount of energy involved; the assigned values were 6, 10 and 15 J. We performed 10 repetitions for each configuration, resulting in a total of 60 ablations. Agar-based gel targets were used for this experiment.

Influence of scanning frequency

Surgeons produce incisions by means of scans, i.e. movements of the laser beam along a trajectory on the surface of the tissue. The robotic platform used in this research (7,11) offers automatic laser scanning, allowing for fast scans at a controlled frequency. Prior research has shown that there is a correlation between the speed of motion of the laser and the resulting depth of incision in hard tissue (3). We wished to determine whether a similar relationship holds true for soft tissue. Therefore, an experiment was performed to determine the influence of the laser scanning frequency on the depth of the resulting incision.

The laser beam was fired on the flat surface of agar-based gel targets while it was automatically scanning along a line of fixed length ($l = 4.6\,\mathrm{mm}$). The scanning frequency was controlled through the scan time, t_s, which defined the time needed to move the beam back and forth along the predefined scan line. Three different values of scan time were considered, 30, 50 and 100 ms. We explored the effect of these speeds at increasing values of exposure time, 3–6 s with increments of 1 s. For each combination of scan time/exposure time, nine repeated trials were executed, resulting in a total of 108 incisions. Constant laser power (3 W) and delivery mode (CW) were used throughout the entire experiment.

Incision depth in *ex vivo* soft tissue

During real interventions, surgeons set the laser parameters (power, delivery mode and scan time) before the execution of an incision. Exposure time is the only input which is manipulated online in order to control the cutting process and the resulting incision depth. To prove the concept of a system able to provide online estimation of incision depth, here we derive a model of incision depth based on the exposure time. Data used to produce this model were collected during experiments involving laser incision of *ex vivo* soft tissue (chicken breast).

We propose that a function f exists, mapping the laser exposure time t_{exp} to the resulting depth of incision, d:

$$d = f(t_{\mathrm{exp}})$$

Incisions were produced to obtain representative input–output data pairs $\left(\left\{t^i_{\mathrm{exp}}, d^i\right\}_{i=1,\ldots,m}\right)$, from which the function f could be estimated.

The input range for these experiments was selected in order to include values of exposure time typically used during real laser microsurgeries. The maximum exposure time in those cases is restricted by the fact that long exposures can produce thermal damage to the tissue (12,13). Surgeons aim to have an exposure time that is long enough for cutting, but still sufficiently short to avoid extensive damage to the surrounding tissue. Values of exposure time were chosen randomly in the selected range (0.5–5 s). A total of 54 incision trials were performed ($P = 3\,\mathrm{W}$, CW, $t_s = 0.1\,\mathrm{s}$, $l = 4.6\,\mathrm{mm}$).

Online estimation of incision depth

Online estimation of cutting depth was integrated in the robotized system for laser surgery of the vocal folds, described in (14). Instead of the micromanipulator used in traditional TLM, this system uses a graphics tablet for laser pointing (Figure 1). This system is equipped with a binocular head-mounted display (HMD), which offers a three-dimensional view of the surgical site.

For this paper, the user interface was customized to display the output of the depth prediction model. A graphical

DOI: 10.1002/rcs

control element was added to the user visualization, showing the progression of the incision depth. This information is represented both numerically (mm) and graphically through the use of a coloured bar gauge (Figure 5). Also, the activation of the laser was done exclusively using the footswitch.

An experiment was performed to assess the effectiveness of the system in supporting precise laser cutting. Three volunteers were involved in this study; they were all members of the scientific staff at our institute and had limited or no prior experience with laser incision of soft tissue. The participants were asked to perform a simple task involving control of the laser exposure time to create a specific depth of incision ($d^* = 0.85$ mm). The incision path was preprogrammed through software and kept constant throughout the experiment. Laser parameters were configured as follows: $P = 3$ W, CW, $t_s = 0.1$ s, $l = 4.6$ mm.

Each participant performed a total of six trials. In the first three trials they were required to accomplish the task relying on their visual perception only. During the three subsequent trials, they were supported by online estimation of depth.

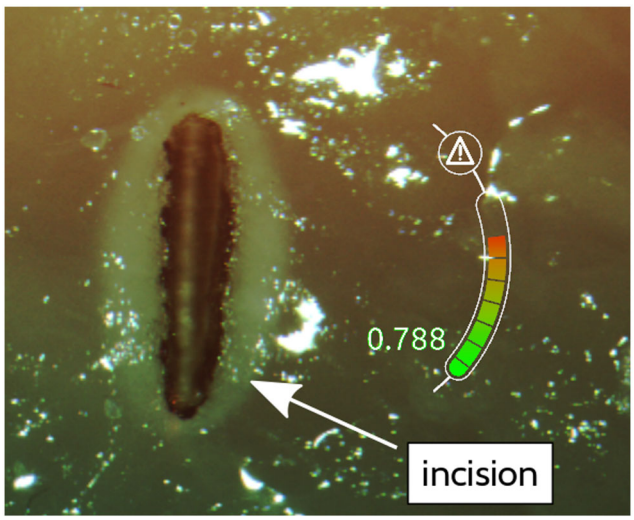

Figure 5. Online estimation of cutting depth. A number indicated the depth of incision (in mm). The same information is represented visually by means of a coloured bar gauge. The top limit of the gauge represents a safety threshold which triggers a warning signal when a certain incision depth is exceeded. The value of the threshold can be configured in the software

Results

Influence of energy delivery mode

Results are presented in Figure 6. As expected, increasing energy densities resulted in deeper ablation craters. The

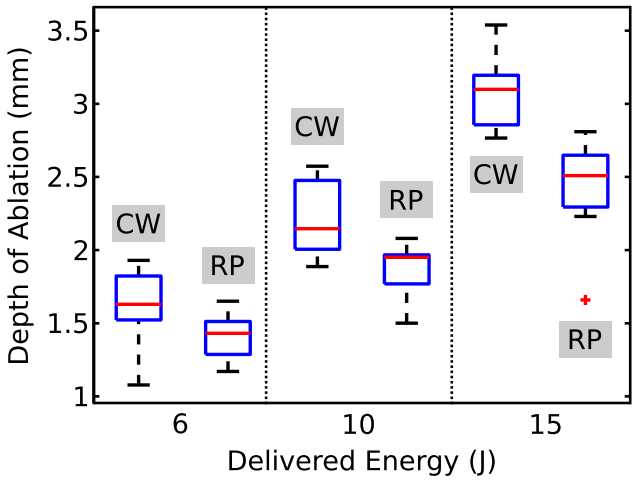

Figure 6. Results of ablation depth produced with different energy delivery modes in agar-based gel targets

experiment revealed that the ablation depth depends not just on the applied energy density but also on the delivery mode. Under fixed energy conditions, the CW mode was found to produce, on average, deeper ablations. The difference between the mean ablation depths obtained in RP and CW was significant for values of energy equal to 10 J ($p = 0.0125$) and 15 J ($p = 0.0011$). In contrast, the difference was not significant at 6 J ($p = 0.0823$).

Table 1 shows the average depth of incision for all the experiments; standard deviation (SD) and coefficient of variation (%) are also listed. Observed variabilities have a marginal significance, as the coefficients of variation are all well below 20%.

Further analysis of experimental data reveals that, once a delivery mode has been fixed, the ablation depth depends linearly on the applied exposure time. A simple linear regression is able to model the relationship between these two quantities (Figure 7) for both CW and RP. In both cases, the fitting error (normalized mean squared error, nMSE) is 0.02%.

Table 1. Mean (d), SD (σ) and coefficient of variation (c_v) of ablation depths produced on agar-based gel targets through different delivery modes

Energy (J)	Delivery mode	d (mm)	σ (mm)	c_v (%)
6	CW	1.619	0.266	16.4
	RP	1.410	0.162	11.4
10	CW	2.215	0.267	12.0
	RP	1.869	0.206	11.0
15	CW	3.075	0.254	8.2
	RP	2.241	0.356	14.7

CW, continuous wave; RP, repeated pulse.

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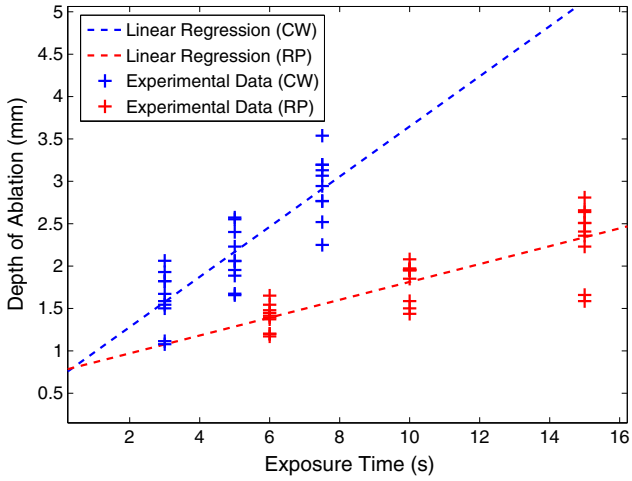

Figure 7. Models of ablation depth in agar-based gel targets for CW (blue) and RP (red) modes

Influence of scanning frequency

Results are shown in Figure 8 [note that one of the targets belonging to the experimental unit (t_{exp} = 6 s, t_s = 30 ms) was accidentally destroyed during manipulation]. Depending on the scan time, different depths of incision were obtained. The collected data show that there is a correlation between scan time and the depth of the resulting incision cavity. Specifically, longer scan times were observed to produce deeper incisions.

Statistics of variability for these experiments are reported in Table 2. With respect to the energy delivery mode experiment, the results are more uniform: observed SDs are one order of magnitude smaller, while the coefficients of variation are all below 13%.

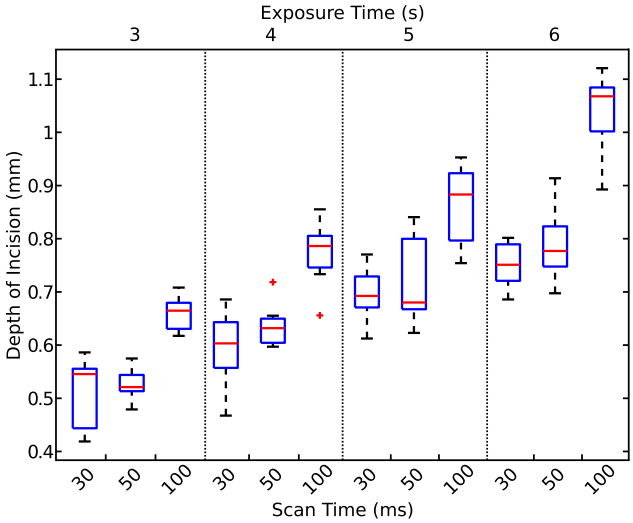

Figure 8. Incision depths produced through different combinations of laser scan time/exposure time in agar-based gel targets

Table 2. Mean (d), SD (σ) and coefficient of variation (c_v) of ablation depths produced on agar-based gel targets through different values of scan time (t_s) and exposure time (t_{exp})

t_{exp} (s)	t_s (ms)	d (mm)	σ (mm)	c_v (%)
3	30	0.50	0.064	12.91
	50	0.63	0.037	5.9
	100	0.86	0.073	8.54
4	30	0.52	0.028	5.4
	50	0.77	0.056	7.32
	100	0.75	0.041	5.6
5	30	0.65	0.03	4.57
	50	0.69	0.045	6.69
	100	0.78	0.065	8.4
6	30	0.59	0.066	11.27
	50	0.72	0.081	11.27
	100	1.04	0.072	6.99

The combined influence of scan time and exposure time on the depth of incision is presented in Figure 9. A simple linear regression of data fits the dataset with a nMSE of 0.6%. A Gaussian mixture regression (15) was also evaluated, resulting in a fitting error of 0.8%.

Incision depth in *ex vivo* soft tissue

The results are shown in Figure 10. It can be seen that the relationship between laser exposure time and incision depth is linear in this region of the input space. Collected data were randomized and divided into a training set and a validation set of 28 and 26 data points, respectively. A simple least square minimization produced a linear estimation of the training set with RMSE = 0.14 mm. The estimated rate of change of the depth was 0.28 mm/s for the specific laser parameters used. The accuracy of this model was evaluated on the validation set, where RMSE = 0.10 mm was obtained.

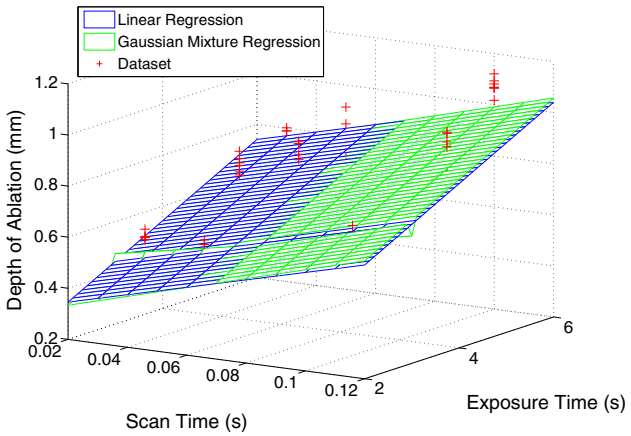

Figure 9. Models of incision depth in agar-based gel targets for different combinations of scan time/exposure time

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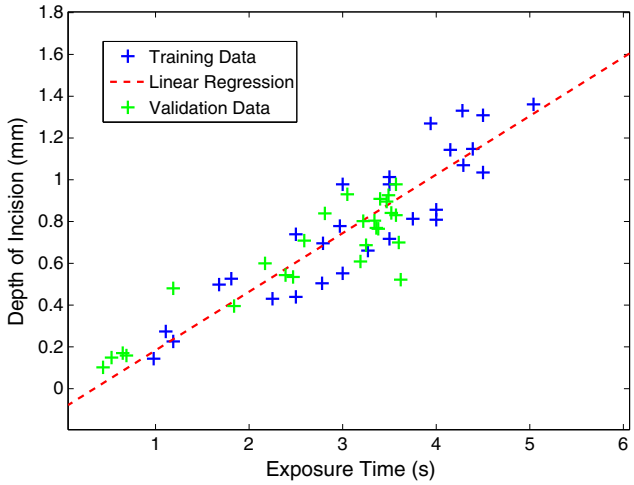

Figure 10. Depths of incision in *ex vivo* soft tissue for different exposure times

Online estimation of incision depth

A comparison of the incision depths produced with and without the support of the online estimation system is shown in Figure 11. None of the users obtained incisions closer than 0.25 mm to the assigned target depth (0.85 mm) during the non-assisted trials. Conversely, with the support of the online estimation system, users produced incisions much closer to the assigned target. Deviations from the target depth are summarized in Table 3.

Discussion

The collected evidence shows that online estimation of laser cutting depth in soft tissue can be enabled by a

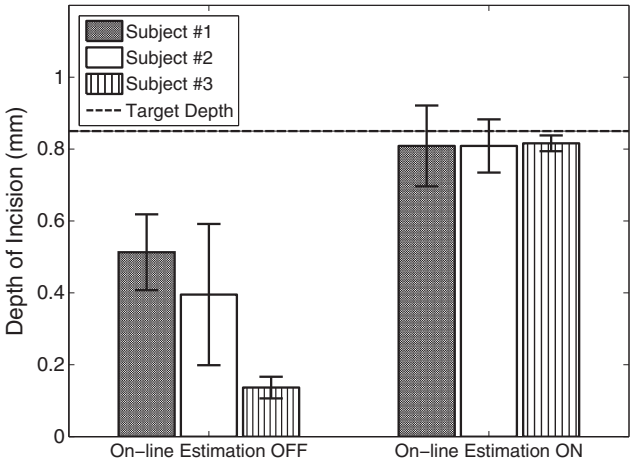

Figure 11. User control of incision depth, with and without the support of the online estimation system. Measures of spread (SD) are graphically represented for each sequence of trials

Table 3. Deviations (RMSE) from the assigned incision depth obtained by users, with and without the assistance of online depth estimation (values in mm)

User	Unassisted	Assisted
1	0.34	0.10
2	0.48	0.07
3	0.71	0.03

function that maps the laser exposure time to the resulting ablation depth. Such a function was extracted from experimental data. The described methodology allowed the creation of a model able to estimate the laser incision depth in *ex vivo* chicken muscle tissue. A linear regression was found to adequately model the relationship between exposure time and the laser incision depth. The model achieved a validation accuracy of 0.1 mm over incisions up to 1.4 mm deep produced on fresh *ex vivo* tissue. The quality of the depth measurements was found to be a crucial aspect of the modelling methodology. Errors in the data can affect the function approximation task, producing inaccurate models. Although different from other methods found in the literature, e.g. confocal microscopy (4,16), optical coherence tomography (17) and inline coherent imaging (5), the protocol described in this paper provides a good resolution for the measurement of incision depths in the range of tenths of millimeters.

The proposed estimation method does not require any additional sensing device, thus it is appropriate for TLM. Considering the resection margins typically employed during TLM, we observe that the accuracy of the model is compatible with the requirements of these interventions. Surgeons aim to reach a minimum of 5 mm in resection depth to achieve surgical radicality (18,19), i.e. to ensure the removal of the whole tumour. Smaller margins, down to 1 mm, are used in those cases where function preservation is considered, including the treatment of glottic cancer (18,19). However, it is important to point out that the reported accuracy was obtained on muscle tissue. Although it was used in the development of the proof-of-concept system, this type of tissue is not representative of the variety of tissues that are encountered during TLM, e.g. epithelium, muscle, adipose and fibrous tissues. These tissues present different optical properties (20), resulting in different laser absorption characteristics. This may mean that the ablation rates (mm/s) of these tissues differ from the linear relationship that we have reported in this study for muscle tissue. The implementation of online estimation of incision depth in a real-case TLM scenario will require the availability of models able to account for the behaviour of different types of tissues. Based on these observations, further experimental work is required to study the ablation rate of different tissues and to find appropriate regression models.

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Online learning methods will be evaluated to enable the improvement of models through the addition of new data.

All laser trials reported here present some degree of variability in the results, which can negatively affect the accuracy of the online depth estimation. This phenomenon has been observed in similar studies (3,4,16) and can be attributed to different factors. For example, instabilities of the laser source may affect the output power of the beam, producing deviations from intended values. Another limiting factor is represented by the non-homogeneous composition of biological tissue. These alterations influence the thermal interaction between laser radiation and tissue (8), thereby hindering the repeatability of laser incisions. However, the levels of variability observed in this study were sufficiently small not to affect the reliability of the depth estimation. Errors observed during the *ex vivo* incision trials support the conclusion that the system provides reliable estimations despite the mentioned repeatability issues.

At the present time, no other real-time assistive system exists to monitor and inform surgeons about the effects of the laser on tissues during the execution of incisions. Cutting trials provided evidence that online estimation of depth is beneficial to users who have no prior experience with laser operations. Providing depth estimations during the execution of laser incisions allowed inexperienced users to perform precise tissue cutting. Our future research efforts will be orientated towards the evaluation of the online estimation of laser cutting depth in a clinical environment.

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Conflict of interest

The authors declare no conflicts of interest.

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References

1. Rubinstein M, Armstrong W. Transoral laser microsurgery for laryngeal cancer: a primer and review of laser dosimetry. *Lasers Med Sci* 2011; **26**: 113–124.
2. Steiner W, Ambrosch P. *Endoscopic Laser Surgery of the Upper Aerodigestive Tract: With Special Emphasis on Cancer Surgery*. Thieme: Stuttgart, New York, 2000.
3. Stopp S, Svejdar D, von Kienlin E, *et al.* A new approach for creating defined geometries by navigated laser ablation based on volumetric 3D data. *IEEE Trans Biomed Eng* 2008; **55**: 1872–1880.
4. Kahrs LA, Burgner J, Klenzner T, *et al.* Planning and simulation of microsurgical laser bone ablation. *Int J Comput Assist Radiol Surg* 2010; **5**: 155–162.
5. Leung BY, Webster PJ, Fraser JM, *et al.* Real-time guidance of thermal and ultrashort pulsed laser ablation in hard tissue using inline coherent imaging. *Lasers Surg Med* 2012; **44**: 249–256.
6. Bay E, Deán-Ben XL, Pang GA, *et al*. Real-time monitoring of incision profile during laser surgery using shock wave detection. *J Biophoton* 2015; **8**: 102–111.
7. Mattos LS, Dagnino G, Becattini G, *et al*. A virtual scalpel system for computer-assisted laser microsurgery. In Proceedings of the 27th IEEE/RSJ International Conference on Intelligent Robots and Systems; 1359–1365.
8. Niemz M. *Laser–Tissue Interactions*. Springer: Berlin, Heidelberg, 2003.
9. Rastegar S, van Gemert MJC, Welch AJ, *et al.* Laser ablation of discs of agar gel. *Phys Med Biol* 1988; **33**: 133–140.
10. Shabihkhani M, Lucey GM, Wei B, *et al.* The procurement, storage, and quality assurance of frozen blood and tissue biospecimens in pathology, biorepository, and biobank settings. *Clin Biochem* 2014; **47**: 258–266.
11. Mattos LS, Deshpande N, Barresi G, *et al.* A novel computerized surgeon–machine interface for robot-assisted laser phonomicrosurgery. *Laryngoscope* 2014; **124**: 1887–1894.
12. Fichera L, Pardo D, Mattos LS. Modeling tissue temperature dynamics during laser exposure. In *Advances in Computational Intelligence: Proceedings of the 12th International Work-conference on Artificial Neural Networks (IWANN 2013), Puerto de la Cruz, Tenerife, Spain, June 2013*, Rojas I, Joya G, Cabestany J (eds). Springer: Berlin, Heidelberg, 2013; 96–106.
13. Fichera L, Pardo D, Mattos LS. Supervisory system for robot-assisted laser phonomicrosurgery. In *Proceedings of the 35th International Conference of the IEEE Engineering in Medicine and Biology Society (EMBC)*, Osaka, Japan, 2013; 4839–4842.
14. Deshpande N, Ortiz J, Caldwell DG, *et al*. A 'virtual microscope' surgeon interface in computer-assisted laser microsurgeries. In *Proceedings of the 2014 International Conference on Robotics and Automation of the IEEE Engineering in Medicine and Biology Society (EMBC)*, Hong Kong, China, 2013.
15. Calinon S. *Robot Programming by Demonstration: A Probabilistic Approach*. EPFL/CRC Press: Lausanne, 2009.
16. Burgner J, Müller M, Raczkowsky J, *et al. Ex vivo* accuracy evaluation for robot assisted laser bone ablation. *Int J Med Robotics Comput Assist Surg* 2010; **6**: 489–500.
17. Kang HW, Oh J, Welch AJ. Investigations on laser hard tissue ablation under various environments. *Phys Med Biol* 2008; **53**: 3381–3390.
18. Hinni ML, Ferlito A, Brandwein-Gensler MS, *et al.* Surgical margins in head and neck cancer: a contemporary review. *Head Neck* 2013; **35**: 1362–1370.
19. Alicandri-Ciufelli M, Bonali M, Piccinini A, *et al.* Surgical margins in head and neck squamous cell carcinoma: what is 'close'? *Eur Arch Otorhinolaryngol* 2013; **270**: 2603–2609.
20. Jacques SL. Optical properties of biological tissues: a review. *Phys Med Biol* 2013; **58**: R37–61.