Kinesthetic and Vibrotactile Haptic Feedback Improves the Performance of Laser Microsurgery

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***Abstract*— The lack of haptic feedback during laser surgery procedures prevents surgeons from accurately discerning the depth of the incisions they perform. In this paper we introduce a novel teleoperated surgical platform, which employs a commercial haptic device to convey information about the laser incision depth to the surgeon. The incision depth is estimated by a feed forward model that maps the laser parameters selected by the surgeon and the total time of laser exposure to the resulting ablation depth. An experiment was conducted to evaluate the effectiveness of the proposed system in enabling precise laser ablation. The experiment involved ten human subjects who were asked to complete a single-point laser ablation task. Results show that haptic feedback can significantly improve the level of surgical precision of laser interventions.**

I. Introduction

Laser surgery is a well-established treatment option for different types of malignancies that affect delicate and small human organs [1]. Transoral Laser Microsurgery (TLM) is one significant example. TLM is a suite of minimally invasive surgical techniques for the management of minuscule laryngeal tumors [2]. In these interventions, a carbon dioxide (CO_2) laser is used as a cutting tool to perform incisions in soft tissue. The goal of TLM is to ensure a complete resection of malignant tissue, as any cancerous cells left in the body may result in a recurrence of the disease [3]. The execution of such accurate tumor resections requires precise control of the laser incisions.

In today's surgical practice, laser incisions are performed manually: surgeons control the laser aiming using a joystick-like device, called laser micromanipulator, while the laser activation/deactivation is controlled with a footswitch. This process does not involve any haptic feedback: the CO_2 laser operates in a contactless (vaporisational) fashion [4], and therefore surgeons cannot use their sense of touch to estimate the depth of the incisions they make, as it would happen if cutting with a scalpel. As a result, surgical precision in TLM procedures largely depends on the dexterity and experience of the operating surgeon. Extensive training is required to develop an effective laser cutting technique, which requires (i) a basic knowledge of the physical principles behind laser ablation of tissues, and (ii) the ability to accurately regulate the laser dosimetry parameters to achieve the desired depth of incision [1]. These laser parameters include laser power, energy delivery mode (continuous or pulsed), pulse duration, and exposure time.

The research leading to these results has received funding from the European Union Seventh Framework Programme FP7/2007-2013 under grant agreement #601165 of project "WEARHAP - WEARable HAPtics for humans and robots", and grant agreement #288233 of project "µRALP - Microtechnologies and Systems for Robot-Assisted Laser Phonomicrosurgery."

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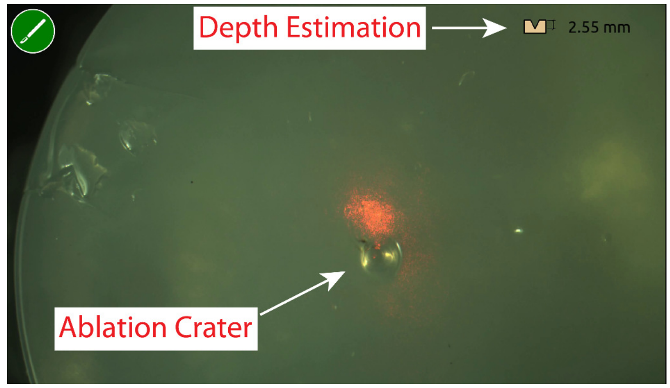


Fig. 1. On-line estimation of the laser cutting depth: a graphical control element is superimposed to a surgical display to inform the surgeon of the depth reached by the laser.

Aiming to overcome these limitations, our research group has recently been working on a novel technology able to estimate the depth of laser incisions during TLM [5]. It uses a model-based supervisor to estimate the laser incision depth in an on-line fashion. Estimates are calculated from the total time of laser exposure and the combination of the laser parameters selected by the surgeon. The estimates are then showed to the surgeon on a display, as illustrated in Fig. 1.

Preliminary experiments have shown that this technique has the potential to enhance surgeons perception of their actions, leading to an improved control of the incision depth [5].

Although effective, visually displaying the depth information may not be the best choice in a real clinical scenario: visual feedback can indeed steer the surgeon's attention from the operating site to the visual feature (e.g. a widget showing the incision depth), increasing the safety risks of the operation. Such risks have been assessed to be significant in different scenarios [6], and it is thus important to avoid them in critical contexts such as surgical procedures. In this respect, haptic feedback may be a safer alternative to visual feedback, providing the necessary information without steering the surgeon's attention away from the operating site.

In this paper we introduce a novel laser microsurgery control interface that uses haptic feedback to provide real-time laser incision depth information to the surgeon. The depth information is rendered to the surgeon through a commercial haptic device, using both kinesthetic and vibrotactile haptic

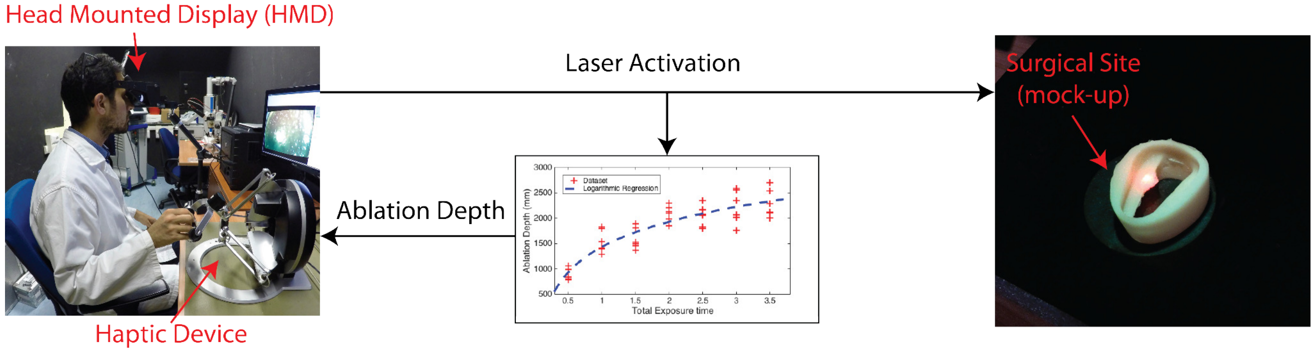


Fig. 2. Block diagram of the proposed haptic system. The surgeon views the surgical site through a stereoscopic display while using the Omega 6 haptic interface to control the laser aiming and its activation. A mathematical model is used to map the total time of laser activation to the resulting laser ablation depth, as summarized in Sec. III-B. This information is rendered to the surgeon through kinesthetic and vibrotactile feedback provided by the Omega haptic interface, as described in Sec. IV.

feedback. Here we report on experimental work aimed at evaluating (i) the level of laser cutting accuracy enabled by the use of haptic feedback, and (ii) the users' confidence in using the proposed system. Furthermore, we provide a comparison with the existing system based on visual feedback and the traditional feedback-less laser cutting method.

II. Related Work

Our work is framed in the field of haptic systems for minimally invasive surgery. This is an active area of study, with numerous recent publications reporting on the use of haptic feedback to address practical problems in the surgical domain. Prior research has established that haptic feedback has the potential to enhance surgical performance in terms of task completion time [7], [8], accuracy [9], [10], [11], peak and mean applied force [8], [12], [13], and fine tool positioning [14]. Studies have also linked the lack of haptic feedback to increased intraoperative injury in minimally invasive surgery operations [15] and endoscopic surgical interventions [16]. Cutaneous haptic feedback has also been recently investigated in the surgical context. Delivering ungrounded cutaneous cues to the human operator conveys in fact information about the forces exerted at the slave side and does not affect the stability of the control loop [17]. For example, Prattichizzo *et al.* [9] showed that cutaneous feedback provided by a moving platform is more effective than sensory substitution via visual feedback in a needle insertion task, and Meli *et al.* [13] found the same type of cutaneous feedback more effective than sensory substitution via either visual or auditory feedback in a pick-and-place task similar to the da Vinci Skills Simulator's Pegboard task. Furthermore, Schoonmaker and Cao [18] and, more recently, McMahan *et al.* [19] demonstrated that vibrotactile stimulation is a viable substitute for force feedback in minimally invasive surgery.

Despite its expected clinical benefits, the use of haptic feedback in laser surgery has not been explored in any significant body of work. A first attempt has been presented by Rizun *et al.* [20]. The authors developed a system that enabled the human operator to feel surfaces using a laser beam, based on optical distance measurements. However, only kinesthetic feedback is evaluated and no application in a real surgical task is considered. The work presented in this paper extends this first attempt by introducing kinesthetic and vibrotactile haptic feedback for a specific surgical application.

III. System Description

We build our system on top of the experimental laser surgery platform called µRALP system [21], which is described in detail in the next subsection. In the scope of this work, the µRALP system was modified by replacing the graphics tablet with a 6-DoF Omega 6 grounded haptic interface (Force Dimension, CH). As we shall see later in this section, this device is used to simultaneously (i) control the laser aiming while (ii) providing feedback on the laser penetration depth into the tissue. Fig. 2 illustrates the setup and the overall architecture of the proposed system.

A. The µRALP Surgical System

The "µRALP Surgical System" [21] is a computer-assisted surgical platform for TLM. This system uses a graphics tablet and stylus for the control of the laser aiming. The commands imparted by the surgeon through this device are mapped into corresponding laser beam trajectories on the surgical site, thus realizing an intuitive (writing-like) control of the laser position. The controlled motion of the laser is provided by a microrobotic system, based on a fast steering mirror [22]. The surgeon also utilizes a footswitch to control the CO_2 laser activation and a stereoscopic display (HMD) to observe the surgical site. This latter device shows the stereoscopic view captured by two HD cameras (Prosilica GT1910 GigE Vision), which are attached to a stereo operating microscope (Leica M651 with xenon lamp, 16x magnification). The HMD is also used to display relevant information to the surgeon. This is the case of the laser ablation depth, which is displayed through the graphical control element shown in Fig. 1.

B. Estimation of the Laser Ablation Depth

The estimation of the laser ablation depth is calculated via software, using the method described in [5]. Estimates are based on a mathematical model that maps the laser parameters and the exposure time to the resulting ablation depth. Such

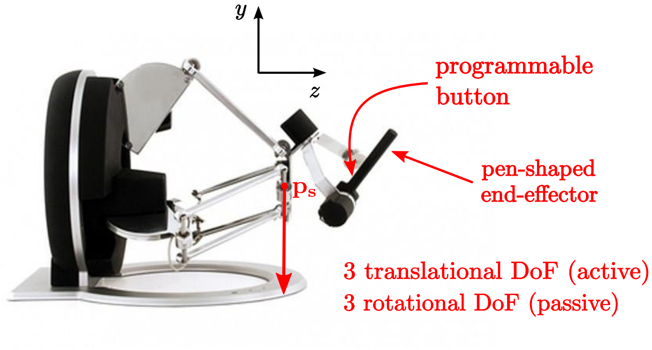


Fig. 3. The Omega 6 haptic interface provides the human operator with information about the laser incision depth, as estimated by the model described in Sec. III-B. The haptic feedback provided is a combination of kinesthetic and vibrotactile stimuli, depending on the feedback modality considered (see Sec. V).

a model is based on statistical regression analysis, and it is generated using data captured during several real laser incisions. Results presented in [5], [23] show that this method enables a level of accuracy up to a tenth of millimiter.

C. Haptic Device

The graphics tablet used with the µRALP system is replaced here by a 6-DoF Omega 6 haptic interface, shown in Fig. 3. Similarly to the graphics tablet, the Omega 6 presents a stylus-like end-effector. However, while the graphics tablet only works in 2D, the Omega 6 presents a delta-based parallel kinematics structure that enables high-accuracy tracking of the end-effector position and orientation in 3D. Translational degrees of freedom are active, while rotational degrees of freedom are passive. The end-effector is equipped with a programmable button. In this work, we use one degree of freedom to provide feedback about the laser incision depth and the programmable button to activate/deactivate the surgical laser and start the tissue ablation. The proposed force feedback technique is described in detail in the next section.

IV. Haptic Rendering of the Laser Ablation Depth

Although the Omega 6 haptic interface can be used to control the laser position in 2D, here – without any loss of generality – we restrict the laser position to a single point. Since the laser ablation is performed at one point, the translational motion of the Omega is constrained on a line, parallel to the y axis (see Fig. 3).

At the beginning of a laser ablation action, the position of the Omega is fixed at point $\mathbf{p_s} = [p_{s,x}\ p_{s,y}\ p_{s,z}]$, and the force provided by the Omega is defined as

$$\mathbf{f(t)} = -k_h(\mathbf{p(t)} - \mathbf{p_s}) - b_c\dot{\mathbf{p}}(\mathbf{t}), \qquad (1)$$

where $k_h = 2000$ N/m, $b_i = 50$ Ns/m, and $\mathbf{p(t)} = [p_x(t)\ p_y(t)\ p_z(t)]$ is the current position of the end-effector of the Omega. The user therefore feels a resistive force when trying to move the end-effector of the Omega away from $\mathbf{p_s}$. The Omega's end-effector is equipped with a programmable button. When this button is pressed, the laser is activated and the ablation starts. As the laser cuts through the tissue, the estimation method previously described in Sec. III-B is used to compute the incision depth $d_e(t)$. The end-effector of the Omega interface moves accordingly along its y axis,

$$f_y(t) = -k_h(p_y(t) - p_{y,d}(t)) - b_c\dot{p}_y(t), \qquad (2)$$

where $p_y(t)$ is the current position of the end-effector of the Omega along the y direction, and $p_{y,d}(t) = p_{s,y} - 16d(t)$ is the current estimated incision depth. The end-effector of the Omega therefore moves down as the laser cuts through the tissue phantom. The scaling factor of 16 between the workspace of the Omega and the estimated incision depth $d_e(t)$ has been chosen to match the magnification ratio of the microscope (see Sec. III-A). Forces along the other two axis, x and z, are computed as in eq. (1), in order to constraint the Omega's motion along its y axis (see Fig. 3).

As soon as the target depth d_t has been reached, i.e. $d_e(t) = d_t$, a 50-ms-long vibrotactile burst is provided to the human operator,

$$\mathbf{f_v(t)} = k_v \begin{bmatrix} \sin(2\pi f_v t) \\ 0 \\ \sin(2\pi f_v t) \end{bmatrix}, \qquad (3)$$

where $k_h = 500$ N/m and $f_v = 150$ Hz. The frequency of vibration is chosen to maximally stimulate the Pacinian corpuscle receptors [24], be easy to distinguish [25], and fit the master device specifications.

As soon as the user release the button, the laser is deactivated and the ablation stops. The task is now considered completed, and the Omega interface returns to its starting position $\mathbf{p_s}$, ready for the next trial.

V. Experimental Evaluation

We carried out experiments with human subjects to evaluate the level of ablation accuracy enabled by the proposed system. The experiment involved volunteer participants, who were asked to complete a simple ablation task. The task consisted in creating ablation cavities with predetermined target depths.

A. Laser Source

The laser source used for this experiment was a Zeiss Opmilas CO_2 25 (wavelength 10.6 µm, TEM_{00} beam profile), configured with the following parameters: 3W laser power, Continuous Wave (CW); 250 μm beam radius. It should be noted that the CW/long-pulsed laser source used here has been superseded in clinical practice by short-pulsed (millisecond) lasers, which are known to produce more efficient cutting and reduced thermal damage [2]. However, using this equipment does not limit the applicability of the illustrated methodology.

B. Tissue Targets

The experiments used cylindrical agar-based gel targets 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. Although different from tissue, these gels offer a convenient and inexpensive medium for laser-tissue interaction studies [26].

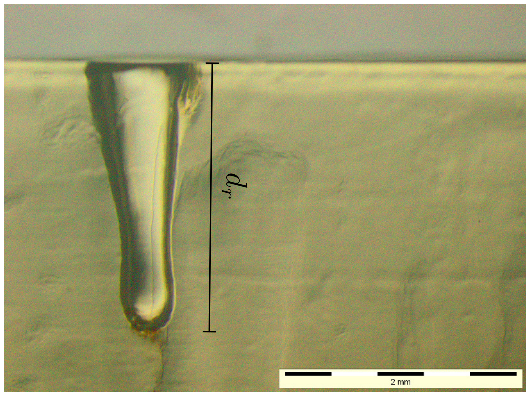


Fig. 4. Transverse profile of an ablation produced on an agar-based gel target. The depth of ablation d_r is estimated by contrasting its size in pixels against the reference scale bar.

C. Experimental Protocol

Ten subjects (8 males, 2 female, age range 26 - 34 years) took part in the experiment, all of whom were right-handed. Eight of them had previous experience with haptic interfaces. None reported any deficiencies in their perception abilities. The experimenter explained the procedures and spent about two minutes adjusting the setup to be comfortable before the subject began the experiment. Before the beginning of the experiment, a 3-minute familiarization period was provided to acquaint them with the experimental setup.

The task consisted of ablating the tissue at three predetermined target depth: 1.5 mm, 1.8 mm, and 2.5 mm. Subjects were informed about the target depth before the beginning of each ablation trial. Each subject made twelve randomized repetitions of the laser ablation task, with three repetitions for each feedback condition proposed (one per each target depth):

- no information on the ablation depth (condition N);
- visual feedback on the estimated ablation depth (condition Vs);
- kinesthetic feedback on the estimated ablation depth (condition K);
- kinesthetic and vibrotactile feedback on the estimated ablation depth (condition K+Vb);

In condition N, subjects are required to activate/deactivate the laser source through the footswitch, as described in Sec. III-A, and no information on the estimated target depth is provided. The Omega interface is not used.

In condition Vs, subjects are again required to activate/deactivate the laser source through the pedal, as described in Sec. III-A. However, in this condition, a visual overlay indicating the estimated target depth is provided, as shown in Fig. 1. The Omega interface is again not used.

In condition K, subjects are required to activate/deactivate the laser source through the Omega's programmable button. As soon as the laser is activated, the Omega's end-effector moves down, as described in Sec. III-C and eq. (2). In this condition, no vibration burst is provided to the subjects when reaching the target depth, i.e. $\mathbf{f_v(t)} = \mathbf{0}$.

In condition K+Vb, subjects are required to activate/deactivate the laser source through the Omega's programmable button. As soon as the laser is activated, the Omega moves down, as for condition K. However, this time a vibration burst is provided to the subject when reaching the target depth, as detailed at the end of Sec. III-C and eq. (3).

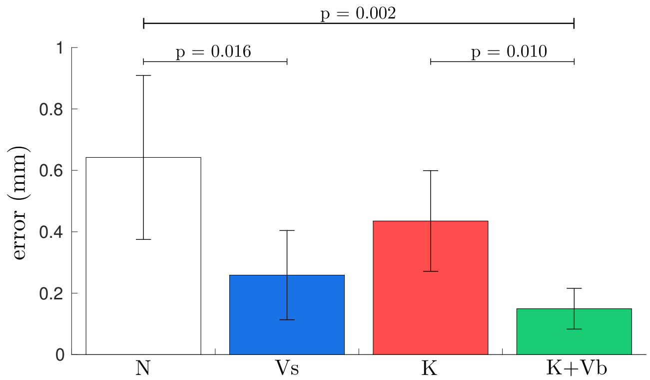
Subjects performed all three repetitions of a single feedback condition as a block, and the order of the conditions and target depths was randomized. Visual feedback on the operating environment was always provided to the subjects through the stereoscopic vision system described in Sec. III-A. The estimation of the incision depth was evaluated according to the algorithm described in Sec. III-B. A video of the experiment is available as supplemental material.

As a measure of performance, we evaluated (1) the real incision error e_r, (2) the estimated incision error e_e, and (3) the error in the estimation of the incision depth e_m. The real incision error e_r is the absolute value of the difference between the incision depth at the end of the task d_r and the target depth d_t indicated to the subject, i.e. $e_r = |d_r - d_t|$. The depth of the incision d_r is measured using a digital microscope (Olympus SZX16). In order to obtain a complete exposure of the crater profile, the agar targets were sectioned into slices. The depth of the incision is defined as the distance from the surface to the bottom of the incision crater, as shown in Fig. 4. 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. The estimated incision error e_e is the absolute value of the difference between the estimated incision depth at the end of the task d_e and the target depth d_t indicated to the subject, i.e. $e_e = |d_e - d_t|$. The depth of the incision d_e is estimated according to the algorithm described in Sec. III-B. The error in the estimation of the incision depth e_m is the absolute value of the difference between the estimated incision depth at the end of the task d_e and the real incision depth d_r measured through the microscope, i.e. $e_m = |d_e - d_r|$. The first two metrics are therefore a measure of performance of the task, while the third one is a measure of performance of our estimation algorithm. A low value of these three metrics denotes the best performance.

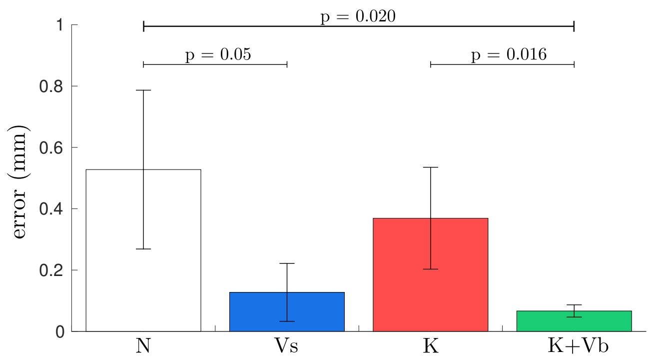
VI. Results

Figure 5a shows the real incision error e_r for the four experimental conditions. All the data passed the Shapiro-Wilk normality test and the Mauchly's Test of Sphericity. A repeated-measure ANOVA showed a statistically significant difference between the means of the four feedback conditions (F(3,27) = 14.029, $p < 0.001$, partial $\eta^2 = 0.609$, a = 0.05). Post hoc analysis with Bonferroni adjustments revealed a statistically significant difference between conditions N and Vs ($p = 0.016$), N and K+Vb ($p = 0.002$), and K and K+Vb ($p = 0.010$). The Bonferroni correction is used to reduce the chances of obtaining false-positive results when multiple pair-wise tests are performed on a single set of data.

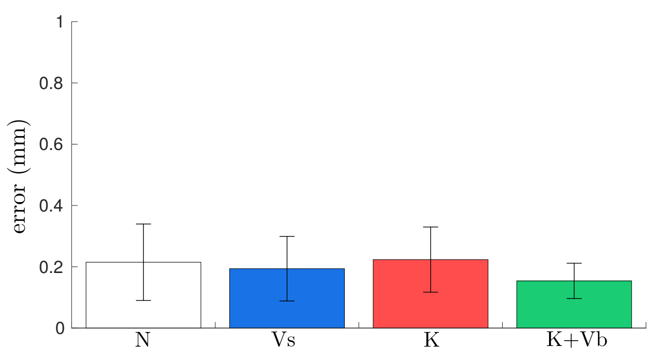
Figure 5b shows the estimated incision error e_e for the four experimental conditions. The collected data passed the



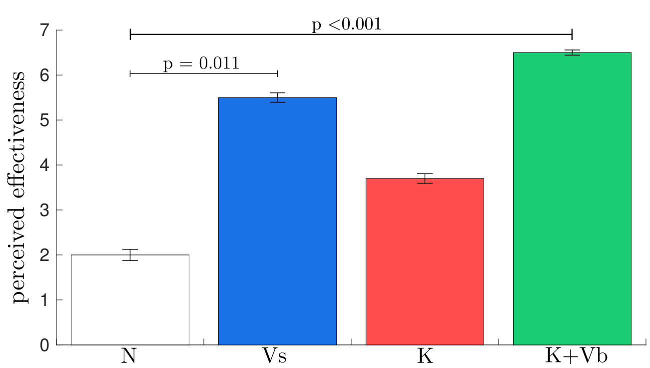
(a) Real incision error



(b) Estimated incision error



(c) Estimation error



(d) Perceived effectiveness

Fig. 5. Experimental evaluation. Real incision error (a), estimated incision error (b), estimation error (c), and perceived effectiveness (d) are evaluated for the condition providing no information about the estimated incision depth (N), visual feedback about the estimated incision depth (Vs), kinesthetic feedback about the estimated incision depth (K), and kinesthetic and vibrotactile feedback about the estimated incision depth (K+Vb). Mean and 95% confidence interval are shown.

Shapiro-Wilk normality test. Mauchly's Test of Sphericity indicated that the assumption of sphericity had been violated ($\chi^2(5) = 17.499, p = 0.004$). A repeated-measures ANOVA with a Greenhouse-Geisser correction showed a statistically significant difference between the means of the four feedback conditions (F(1.568,14.109) = 8.982, $p < 0.001$, partial $\eta^2 = 0.500$, a = 0.05). Post hoc analysis with Bonferroni adjustments revealed a statistically significant difference between conditions N and Vs ($p = 0.05$), N and K+Vb ($p = 0.020$), and K and K+Vb ($p = 0.016$).

Figure 5c shows the error in the estimation of the incision depth e_m for the four experimental conditions. All the data passed the Shapiro-Wilk normality test and the Mauchly's Test of Sphericity. A repeated-measure ANOVA showed no statistically significant difference between the means of the four feedback conditions (F(3,27) = 0.731, $p = 0.543$, partial $\eta^2 = 0.075$, a = 0.05).

In addition to the quantitative evaluation reported above, we also measured users' experience. Immediately after the experiment, subjects were asked to report the effectiveness of each feedback condition in completing the given task using bipolar Likert-type seven-point scales. Figure 5d shows the perceived effectiveness of the four feedback conditions. A Friedman test showed a statistically significant difference between the means of the four feedback conditions (χ^2(3) = 20.463, $p < 0.001$). The Friedman test is the non-parametric equivalent of the more popular repeated measures ANOVA. The latter is not appropriate here since the dependent variable was measured at the ordinal level. Post hoc analysis with Bonferroni adjustments revealed a statistically significant difference between conditions N and Vs ($p = 0.011$), and N and K+Vb ($p < 0.001$). Finally, subjects were asked to choose the condition they preferred the most. Condition K+Vb was preferred by eight subjects and condition Vs was preferred by two subjects. All the subjects reported that during condition Vs were not able to focus on the ablation point, but they were only looking at the visual overlay. On the other hand, during condition K+Vb they were able to wait for the vibration burst while looking at the ablation.

VII. Discussion

Providing depth information via kinesthetic and vibrotactile feedback (K+Vb) or with a visual gauge (Vs) significantly outperformed the condition providing no feedback at all (N) in all the considered metrics. Moreover, although the improvement was not found statistically significant, the condition providing kinesthetic and vibrotactile feedback (K+Vb) outperformed the condition providing visual feedback (Vs) in all the considered metrics. In addition, kinesthetic feedback alone (K) seems to reduce incision error and its variability with respect of the control condition (N), albeit non-significantly. Also, vibrotactile feedback (K+Vb) further improved such measure in a significant manner.

Condition K+Vb was also preferred by 80% of the subjects, who particularly appreciated the capability of condition K+Vb to enable them to focus on the ablation point. During condition Vs, in fact, subjects reported to focus only on the visual feedback provided by the system and not on the ablation point. Although this fact was not an issue in our experimental evaluation, since the laser beam was not moved during the experiment, it may play a significant role if the operator is required to move the laser along a trajectory to realize incisions. This issue could be addressed through a redesign of the visual feedback modality, e.g. moving the visual feature closer to the ablation point. Indeed, any redesign should be carefully planned not to obstruct the surgeon's view of the surgical site. In this perspective, haptic feedback presents the advantage of using a sensory channel that is currently not utilized, eliminating the need for visual cues and enabling the surgeon to focus his visual attention on the surgical site.

These results seem to indicate that the use of haptic feedback has the potential to increase the accuracy of laser microsurgery. However, it is important to point out that only volunteers were involved at this stage, none of whom was a surgeon and the majority of whom (8 out of 10) had prior experience with haptic interfaces. To further ascertain the validity of the approach presented here, we plan to conduct a new experimental campaign, this time involving surgeons, as they are the end users.

VIII. Conclusions and Future Work

This paper presented a novel teleoperation system for laser incision in soft tissue. The system uses kinesthetic and vibrotactile haptic feedback to inform the operator of the estimated laser incision depth. Experimental data suggests that this system has a potential to increase the precision of laser microsurgery interventions.

Future efforts will be directed at validating this system with the involvement of laser surgeons. The ultimate goal consists in the realization of an interface that will enable clinicians to simultaneously control the length, shape, and depth of a laser incision.

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