# Automatic Focus Adjustment for Single-Spot Tissue Temperature Control in Robotic Laser Surgery

Nicholas E. Pacheco, Chaitanya S. Gaddipati, Siavash Farzan, Loris Fichera

***Abstract*—This paper reports on a study whose goal is to control the tissue temperature at a specific spot during laser surgery — for the purpose of, e.g., inducing coagulation or sealing blood vessels. We propose a solution that relies on the automatic adjustment of the laser focus (and thus how concentrated the laser beam is), combined with the use of an infrared thermal camera for non-contact temperature monitoring. One of the main challenges in the control of thermal laser-tissue interactions is that these interactions can be hard to predict due to the inherent variability in the molecular composition of biological tissue. To tackle this challenge, we explore two different control approaches: (1) A model-less controller using a Proportional-Integral (PI) formulation, whose gains are set via a tuning procedure performed on laboratory-made tissue phantoms; and (2) A model-based controller using an *adaptive* formulation that makes it robust to tissue variability. We report on experiments, performed on four types of tissue specimens, showing that both controllers can consistently achieve temperature tracking with a Root-Mean-Square Error (RMSE) of $\approx 1\,^{\circ}\mathrm{C}$.**

***Index Terms*—Surgical Robotics; Laser Surgery; Laser Focus; Laser-Tissue Interactions.**

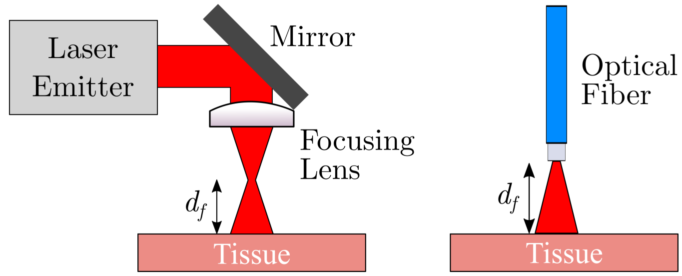


Fig. 1. The goal of laser focusing is to create a spot size of prescribed width via the control of $d_f$, i.e., the distance between the laser beam's focal point and the tissue surface. (Left) In free beam systems, the location of the focal point depends on the characteristics of the lenses used to focus the beam. (Right) In fiber-based systems, laser light diverges immediately upon exiting the fiber, with an angle determined by the numerical aperture of the fiber itself. In this manuscript, we study how regulating $d_f$ can be used to produce controlled single-spot tissue heating.

## I. Introduction

LASERS are an important tool in modern medical practice. Within the context of surgery, lasers are frequently used as a cutting instrument, e.g., to excise tumors. Several research groups have recently developed robotic systems for laser surgery [1]–[7], with the goal of providing enhanced laser aiming and cutting precision. Within this area of research, one of the problems that has received considerable attention is the automatic control of the laser focus. Briefly, laser focusing refers to the process of optically adjusting a laser beam so that it is concentrated in a small, well-defined spot – see Fig. 1. In surgical applications, tight laser focusing is desirable to maximize cutting precision; yet, focusing can be hard to perform manually, as even slight variations ($<$ 1 mm) in the focal distance can significantly affect the spot size. Motivated by these challenges, Kundrat and Schoob [8], [9] recently introduced a technique to robotically maintain constant focal distance, thus enabling accurate, consistent cutting. In another study, Geraldes and colleagues [10] developed an automatic focus control system based on a miniaturized *varifocal* mirror, and they obtained spot sizes as small as $380\,\mu\mathrm{m}$.

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Whereas previous work has mainly dealt with the problem of creating – and maintaining – small laser spots, in this paper we explore the converse problem, i.e., to controllably defocus surgical lasers. In clinical practice, physicians defocus a laser beam whenever they wish to change its effect from cutting to heating [11]–[13]— e.g., to thermally seal a blood vessel. To the best of our knowledge, no previous work has studied the problem of robotically regulating the laser focus to achieve controlled tissue heating, which is precisely the contribution of the present manuscript.

In the following sections, we first briefly review the physics of thermal laser-tissue interactions and then discuss possible approaches to regulate tissue heating according to a prescribed temperature profile. Laser-tissue interactions are generally considered hard to control due to the inherent inhomogeneity of biological tissue, which can create significant variability in its thermal response to laser irradiation [14]. To tackle this challenge, we investigate two distinct approaches: (1) A model-less strategy employing a Proportional-Integral (PI) controller, whose gains are set via a tuning procedure that uses laboratory-made tissue phantoms; and (2) A model-based approach based on an *adaptive* controller, whose formulation makes it robust to tissue variability [15]. We report experimental evidence showing that both of the proposed controllers can achieve accurate temperature tracking on different types of tissue, without requiring prior knowledge of the tissue's physical properties. Benefits and limitations of each control approach are discussed at the end of the paper.

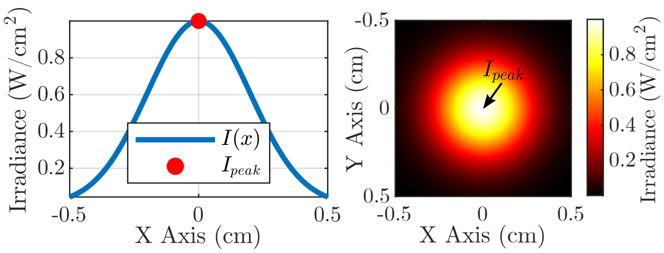


Fig. 2. (Left) 1D and (Right) 2D normalized irradiance profiles for a Gaussian laser beam $I(x, y)$. Light intensity is the strongest at the center of the beam ($I_{\text{peak}}$) and fades radially.

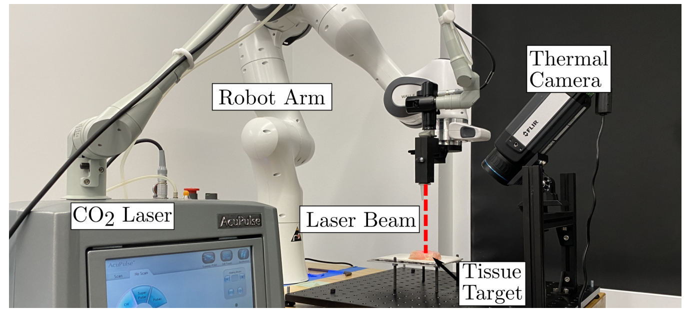


Fig. 3. Experimental Setup.

## II. Materials and Methods

### A. Problem Formulation

Let us consider a scenario where a tissue specimen is exposed to a laser beam of intensity $I$ ($\mathrm{W\,cm^{-2}}$). The problem we wish to solve is to control the temperature around the point of incidence of the laser on the tissue surface. We assume that the laser delivery system can only be moved vertically with respect to the tissue, i.e., that the laser is stationary and the only variable we can control is the distance $d_f$ between the beam's focal point and the tissue surface (refer to Fig. 1).

In general, thermal laser-tissue interactions can be modeled as the result of a heating process, with the addition of a heat dissipation term [1], i.e.,

$$c_v \frac{\partial T(x, y, t)}{\partial t} = \mu_a I(x, y) + k\ \nabla^2 T(x, y, t). \tag{1}$$

Here, $x$ and $y$ are the coordinates of a Cartesian reference system on the tissue surface and $t$ represents time; $T$ is the tissue temperature, and $c_v$, $k$, $\mu_a$ are three tissue-specific physical parameters: Namely, $c_v$ is the *volumetric heat capacity* ($\mathrm{J\,cm^{-3}\,{}^\circ C^{-1}}$), $k$ is the *thermal conductivity* ($\mathrm{W\,cm^{-1}\,{}^\circ C^{-1}}$), and $\mu_a$ is the *coefficient of absorption* of the laser ($\mathrm{cm^{-1}}$). We note that these parameters are rarely known with certainty, as different types of tissue will generally have different physical properties, and significant variations are possible even within specimens of the same tissue type [16], [17]. Our goal in this study is to synthesize controllers able to regulate the dynamics in Eq. (1) without explicit knowledge of these properties.

Most surgical laser systems emit beams with Gaussian-shaped intensity, as illustrated in Fig. 2, and the temperature distribution created by these lasers follows a similar bell-shaped profile. Based on this, we formulate our control objective as the problem of regulating the peak surface temperature $T_{\text{peak}}$, which is normally observed at the center of the beam, where the light intensity reaches its maximum value $I_{\text{peak}}$.

It is possible to control the peak intensity $I_{\text{peak}}$ of the beam by adjusting the focal distance $d_f$ of the laser, using the following model of beam divergence from [1]:

$$d_f = \frac{\pi w^2}{\lambda} \sqrt{\frac{2P}{I_{\text{peak}}\ \pi w^2} - 1}. \tag{2}$$

In the equation above, $w$ is the *beam waist* (i.e., the radius at which the beam intensity fades to $1/e^2$ of its peak value, measured at the focal point), $\lambda$ is the laser wavelength, and $P$ is the laser optical power.

### B. Experimental Setup

To control the tissue temperature, we synthesized two controllers: a PI controller (described in Sec. III below) and an adaptive controller (refer to Sec. IV). We verified each controller's performance with experiments on *ex-vivo* tissue, using the setup shown in Fig. 3. Experiments used a surgical $CO_2$ laser, the Lumenis AcuPulse (Lumenis, Israel), whose beam is delivered through an articulated (passive) arm. The distance $d_f$ between the focal point of the beam and the tissue was controlled by a Panda robotic arm (Franka Emika, Germany). The tissue surface temperature was monitored with an A655sc thermal camera (Teledyne FLIR, Oregon, USA) at a rate of 100 frames per second (fps).

We carried out experiments on four different types of tissue: soft tissue phantoms (agar gelatin) and *ex-vivo* chicken muscle, bovine liver, and bovine bone. The latter three specimens were sourced from a local butcher shop, while the agar tissue phantoms were fabricated in our laboratory using a mixture of 2% agar powder (Sigma-Aldrich Chemie, Germany) and 98% deionized water. In each experiment, we prescribed a temperature profile $T_r(t)$ which first linearly ramps up to 50 °C, then remains constant for 60 seconds. We carried out five repetitions for each tissue type, for a total of 20 experiment runs per controller.

## III. PI Control

As a first attempt to control the tissue heating, we consider a simple PI controller:

$$I_{\text{peak}}(t) = k_p e(t) + k_i \int_0^t e(\tau)\ d\tau. \tag{3}$$

Here, $k_p$ and $k_i$ are positive gains, and the error term $e(t)$ is defined as $e(t) = T_r(t) - T_{\text{peak}}(t)$. The equation above is used in conjunction with Eq. (2) to regulate the focal distance $d_f$.

### A. Controller Tuning

The controller gains $k_p$ and $k_i$ were tuned by applying repeated laser pulses on agar tissue phantoms, and by manually adjusting their values in an attempt to minimize the settling time and overshoot of the thermal response. The rationale for tuning the controller on agar phantoms is that the absorption of infrared laser light (such as the one emitted by our $CO_2$ laser) in biological tissue is primarily driven by the presence of water in the tissue itself. As agar phantoms are primarily made of

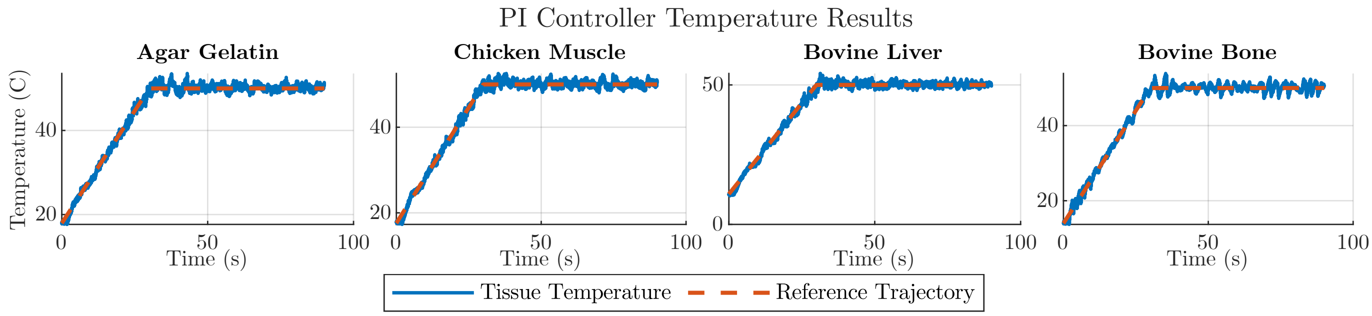


Fig. 4. PI controller results. Of the five trials performed on each tissue type, here we report the ones where we observed the median within-group RMSE.

water, they offer a controlled and convenient medium to mimic the thermal response of biological tissue to laser irradiation. We found that $k_p = 0.3$ and $k_i = 0.008$ provided reasonable tracking accuracy on the agar phantoms.

### B. Controller Performance

Results are shown in Fig. 4 and summarized in Table I. We found the PI controller to be robust to variations in tissue type, with the most accurate tracking performance obtained on chicken muscle (average RMSE across the five trials: 0.74 °C), while the highest error was observed on bovine bone (average RMSE: 1.07 °C).

## IV. Adaptive Controller

While the PI controller exhibits reasonable accuracy, it is fair to wonder whether a model-based controller can provide better performance. To overcome the lack of knowledge of the tissue physical properties in Eq. (1), here we consider model-reference adaptive control (MRAC), i.e., a family of well-known control methods for systems with uncertain or unknown parameters [15]. The control law we propose is:

$$I_{\text{peak}}(t) = \hat{a}_y T_{\text{peak}}(t) + \hat{a}_f \hat{f}(T(x,y,t)) + \hat{a}_r T_r(t), \quad (4)$$

where $\hat{f}(T(x,y,t))$ numerically approximates the dissipation term in Eq. (1). In the equation above, $\hat{a}_y$, $\hat{a}_f$, $\hat{a}_r$ are three *adaptive* parameters whose values are dynamically adjusted over time to minimize the tracking error. In addition to the control law, the synthesis of an MRAC controller requires the specification of a reference model, i.e., a model describing the dynamics that the adaptive controller should seek to create by adjusting $\hat{a}_y$, $\hat{a}_f$, $\hat{a}_r$. Here, we consider the following first-order reference model and corresponding error $e(t)$:

$$\begin{aligned} \dot{T}_m(t) &= -a_m T_m(t) + b_m T_r(t), \\ e(t) &= T_{\text{peak}}(t) - T_m(t), \end{aligned} \quad (5)$$

with $a_m$ and $b_m$ being strictly positive constants. With these definitions, we can formulate the update rules for the adaptive gains as:

$$\begin{aligned} \dot{\hat{a}}_y &= -\gamma_y\, e(t)\, T_{\text{peak}}(t), \\ \dot{\hat{a}}_f &= -\gamma_f\, e(t)\, \hat{f}(T(x,y,t)), \\ \dot{\hat{a}}_r &= -\gamma_r\, e(t)\, T_r(t), \end{aligned} \quad (6)$$

with $\gamma_y, \gamma_f, \gamma_r$ being three adaptive update gains.

TABLE I
THE MEAN (STD) ROOT MEAN SQUARED ERROR (°C) OF EACH CONTROLLER'S TRACKING ACCURACY FOR EACH MATERIAL.

| | Agar | Muscle | Liver | Bone |
|---|---|---|---|---|
| PI Control | 0.81 (0.02) | 0.74 (0.04) | 1.02 (0.15) | 1.07 (0.28) |
| Adaptive Control | 0.79 (0.05) | 0.82 (0.08) | 1.08 (0.31) | 1.89 (0.84) |

### A. Controller Initialization

In general, the tracking accuracy of an adaptive controller can be enhanced if a *reasonable* estimate of the unknown system parameters is available. We initialize our controller based on the physical properties $k$, $c_v$, and $\mu_a$ of the agar tissue phantoms produced in our laboratory (as we have seen in the previous section, agar phantoms provide a medium that can mimic the thermal response of tissue to laser irradiation). The calculations used the following three empirical relations from the literature on laser-tissue interactions [14], [16]:

$$\begin{aligned} c_v &= (1.55 + 2.8w)\rho \\ k &= 0.0006 + 0.0057w \\ \mu_a &= w\mu_{aw} \end{aligned} \quad (7)$$

where $\rho$ is the tissue density ($\frac{\text{g}}{\text{cm}^3}$), $w$ is the water content, and $\mu_{aw}$ is the absorption coefficient of water. The following approximations were obtained: $c_v = 5.11\ \text{J}\,\text{cm}^{-3}\,°\text{C}^{-1}$ and $k = 0.0062\ \text{W}\,\text{cm}^{-1}\,°\text{C}^{-1}$, $\mu_a = 784\ \text{cm}^{-1}$.

### B. Controller Performance

Temperature tracking results are reported in Fig. 5 and Table I. The adaptive controller generally showed robustness to variations in tissue type, with the exception of bone, where we observed a noticeably higher error (average RMSE across the five trials: 1.89 °C).

## V. Discussion

In general, our findings suggest the viability of regulating the tissue temperature at a single spot by adjusting the focal distance of the laser. While the thermal camera used in this work would likely be impractical to use in a minimally invasive surgery setting, we believe that it would be possible to implement the proposed method using a miniaturized non-contact infrared imager, such as the FLIR Lepton (Teledyne FLIR, Oregon, USA).

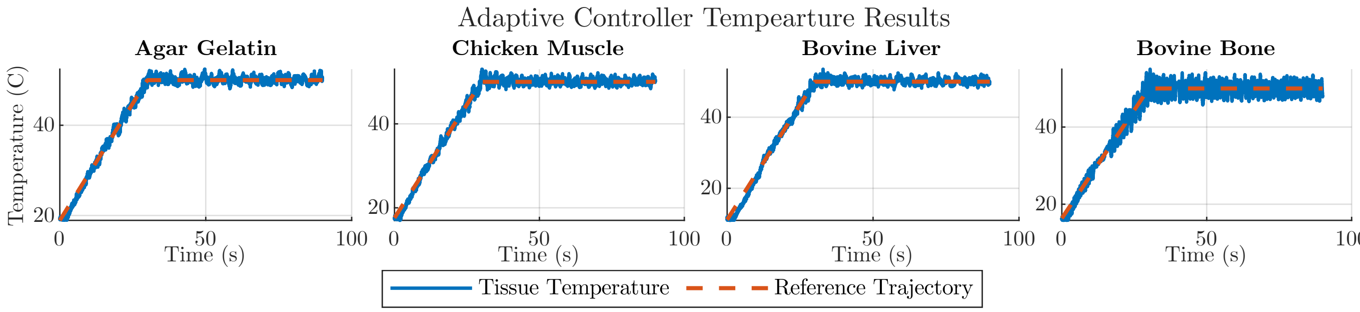


Fig. 5. Adaptive controller results. Of the five trials performed on each tissue type, here we report the ones where we observed the median within-group RMSE.

On soft tissue specimens (agar phantoms, chicken muscle, and bovine liver), both the proposed PI and adaptive controllers were able to consistently achieve temperature tracking with an RMSE of $\approx 1\,^{\circ}\mathrm{C}$ or lower. Both controllers also experienced a performance degradation on bovine tissue, and especially bone, with the adaptive controller (surprisingly) performing significantly worse than the PI controller. The degraded performance of the PI controller could be explained in light of the peculiarity of bone tissue, which is significantly different in composition (and especially water content) than the agar gels used for controller tuning. For what concerns the adaptive controller, we believe that its performance on bone can be attributed to the fact that its control law does not include a proportional term (see Eq. (4)). Lacking such a term, the controller will respond more slowly to an increase in the temperature error, leading to a larger error overall. Additionally, several modeling assumptions made in the formulation of the temperature dynamics (Eq. (1)), which could have affected the performance of the adaptive controller. Among other things, we neglected optical effects such as reflectance, refraction, and scattering which may occur in the tissue. While these effects are often assumed to be negligible in first approximation [14], they can reduce or otherwise alter the intensity of the laser beam, and thus the resulting thermal dynamics. Future studies will have to be conducted to see if accounting for these effects can lead to improved temperature tracking accuracy.

## VI. Conclusion

This paper presented a novel method to implement single-spot temperature control in robotic laser surgery. The proposed method controls tissue heating by robotically regulating the laser focus. We experimentally evaluated and compared two different control approaches, a PI controller, and an adaptive controller, both of which were able to achieve temperature tracking with a Root-Mean-Square Error (RMSE) of $\approx 1\,^{\circ}\mathrm{C}$ on different tissue types.

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