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) $\approx 1\,^{\circ}\mathrm{C}$.**

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

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]–[8], with the goal of providing enhanced laser aiming and patient safety. 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 [9], [10] recently introduced a technique to robotically maintain constant focal distance, thus enabling accurate, consistent cutting. In another study, Geraldes and colleagues [11] 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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N.E. Pacheco, C.S. Gaddipati, and L. Fichera are with the Department of Robotics Engineering, Worcester Polytechnic Institute, Worcester, MA 01609, USA.

S. Farzan is with the Department of Electrical Engineering at California Polytechnic State University, San Luis Obispo, CA 93407, USA.

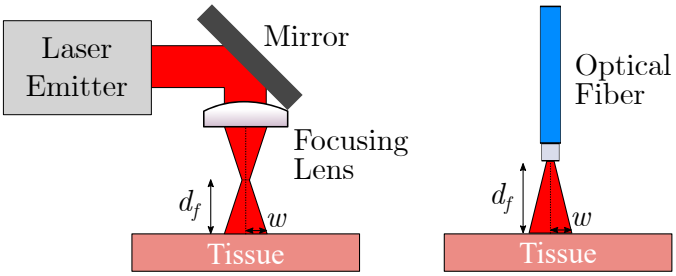

Fig. 1. The goal of laser focusing is to create a spot size of prescribed radial width, w, via the control of d_f, i.e., the distance between the laser beam's focal point and the tissue surface. (Left) In free beam systems, the location of the focal point depends on the optical characteristics of the beam, including the effect of any lenses that may be employed to focus the light. (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.

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 defocus the laser beam in a controlled manner. In clinical practice, physicians defocus a laser beam whenever they wish to change its effect from cutting to heating [12]–[14] 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.

This manuscript is structured as follows: after a review of related work, we briefly describe the physics of thermal laser-tissue interactions and then discuss possible approaches to regulate tissue heating through laser defocusing. 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 [15]. To tackle this challenge, we investigate two distinct approaches: (1) a model-less strategy employing a Proportional-Integral (PI) controller, whose gains are set through a tuning procedure that uses laboratory-made tissue phantoms; and (2) a model-based approach based on an *adaptive* controller [16], whose formulation makes it robust to tissue variability. 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. The benefits and limitations of each control approach are discussed at the end of the paper.

II. Related Work

The problem of modeling and controlling the thermal interactions that occur during robotic laser surgery has not been thoroughly explored until now [1]. Previous research by Pardo, Fichera, and colleagues [17]–[19] sought to use data-driven methods to characterize the thermal response of the tissue, with the overarching goal of building an automatic emergency laser cut-off system, i.e., a system capable of preventing accidental tissue overheating (and thus thermal injuries) [2]. The work we describe in this manuscript extends previous research by exploring, for the first time, technical approaches to robotically regulate the tissue temperature, rather than simply implementing an emergency shut-off.

Outside the scope of robotic surgery, related work has been carried out for the development of laser ablation therapies, particularly Laser Interstitial Thermal Therapy (LITT), a minimally-invasive procedure used to treat tumors not amenable to surgical resection [15]. LITT uses laser light, guided by an optical fiber directly inside a tumor, to thermally destroy diseased tissue from within. Until now, most of the effort in this area of research has focused on the development of temperature sensing technology [20]. Temperature control is typically implemented by modulating the laser intensity with simple On-Off or PID controllers [21]–[23], with control parameters tuned *ad-hoc* for specific tissue types. The work we describe herein seeks to advance our ability to control thermal laser-tissue interactions through the synthesis of controllers that are tissue-agnostic, i.e., controllers able to regulate temperature without requiring prior knowledge of either the tissue type or its physical properties.

III. Materials and Methods

A. Problem Formulation

Consider a scenario where a tissue specimen is exposed in air 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 by regulating the spot size of the beam, i.e., its radial width w, as illustrated in 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, and μ_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 μ_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 [24], [25]. 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_{peak}, which is normally observed at the center of the beam, where the light intensity reaches its maximum value I_{peak}. For a Gaussian beam, I_{peak} is related to the radial width w by the following relation [1]:

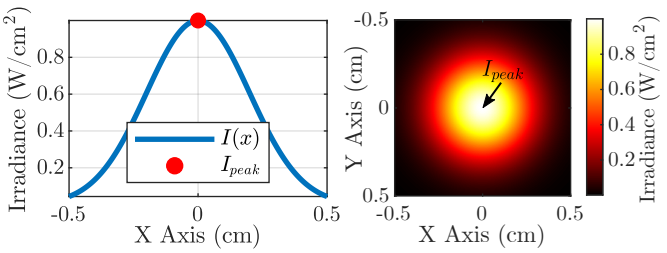

Fig. 2. (Left) 1D and (Right) 2D normalized irradiance profiles for a Gaussian laser beam $I(x,y)$. Light intensity is the strongest at the center of the beam (I_{peak}) and fades radially.

$$I_{\text{peak}} = \frac{2P}{\pi w^2}, \tag{2}$$

where P is the laser optical power.

In this study, the beam's radial width w is controlled by adjusting the working distance d_f, i.e., the distance between the focal point of the laser and the tissue surface (refer to Fig. 1). This is accomplished by moving the laser delivery apparatus closer to (or further away from) the tissue, using a robotic manipulator. The relation between w and d_f is given by the following model of beam divergence [1]:

$$w = w_0 \sqrt{1 + \left(\frac{\lambda d_f}{\pi w_0^2}\right)^2}, \tag{3}$$

where w_0 is the *beam waist* (i.e., the beam's radial width at the focal point) and λ is the laser wavelength. Combining Eqs. (2) and (3) yields the following relation, which we use to control d_f based on a desired value of peak intensity I_{peak}:

$$d_f = \frac{\pi w_0^2}{\lambda} \sqrt{\frac{2P}{I_{\text{peak}}\ \pi w_0^2} - 1}. \tag{4}$$

B. Experimental Setup

To control the tissue temperature, we synthesized two controllers: a PI controller (described in Sec. IV below) and an adaptive controller (refer to Sec. V). We verified each controller's performance with experiments on *ex-vivo* tissue, using the setup shown in Fig. 3.

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 $50\,^\circ\mathrm{C}$, then remains constant for 60 seconds. We carried out

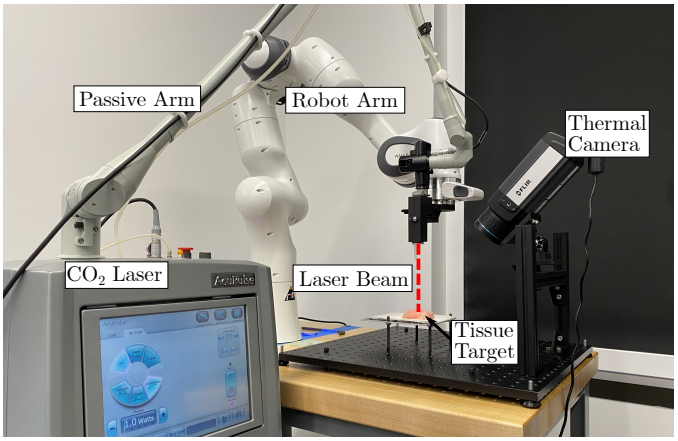

Fig. 3. Experiments used a surgical CO_2 laser, the Lumenis AcuPulse (Lumenis, Israel), whose beam is delivered through an articulated (passive) arm. For this laser, the beam waist w_0 is 0.21 mm. The output power is set through a control panel; all experiments reported in this manuscript used a power level of 1 W. The beam's focus (i.e., the radial width w measured at the tissue surface) was controlled by moving the laser handpiece closer to (or further away from) the tissue surface with 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) and spatial resolution of 60 $\frac{\text{pixels}}{\text{cm}}$. T_{peak} was taken as the pixel recording the highest temperature.

five repetitions for each tissue type, for a total of 20 experiment runs per controller.

IV. PI Controller

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

$$I_{\text{peak}}(t) = k_p e(t) + k_i \int_0^t e(\tau)\ d\tau. \tag{5}$$

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. (4) to regulate the working 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 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 temperature tracking 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).

V. Adaptive Controller

While the PI controller exhibits reasonable accuracy, we wish to investigate whether a model-based controller could provide better performance. To overcome the lack of knowledge of the tissue's 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 [16]. The control law we propose is:

$$I_{\text{peak}}(t) = \hat{a}_y T_{\text{peak}}(t) + \hat{a}_f \hat{f}(T(x,y,t)) + \hat{a}_r T_r(t), \tag{6}$$

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 $e(t)$:

$$\begin{aligned} \dot{T}_m(t) &= -a_m T_m(t) + b_m T_r(t), \\ e(t) &= T_{\text{peak}}(t) - T_m(t), \end{aligned} \tag{7}$$

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

$$\begin{aligned} \dot{\hat{a}}_y &= -\gamma_y\, e(t)\, T_{\text{peak}}(t), \\ \dot{\hat{a}}_f &= -\gamma_f\, e(t)\, \hat{f}(T(x,y,t)), \\ \dot{\hat{a}}_r &= -\gamma_r\, e(t)\, T_r(t), \end{aligned} \tag{8}$$

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

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 μ_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 [15], [24]:

$$\begin{aligned} c_v &= (1.55 + 2.8w)\rho, \\ k &= 0.0006 + 0.0057w, \\ \mu_a &= w\mu_{aw}, \end{aligned} \tag{9}$$

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

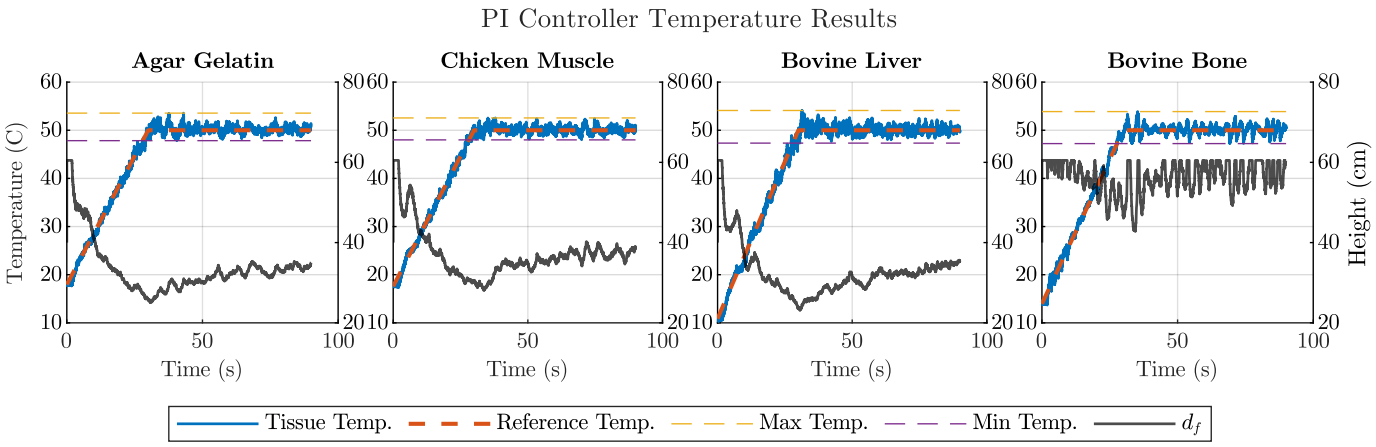

Fig. 4. PI controller results. Of the five trials performed on each tissue type, here we report temperature tracking accuracy (left y-axis) and the robot height setpoint (right y-axis) where we observed the median within-group RMSE.

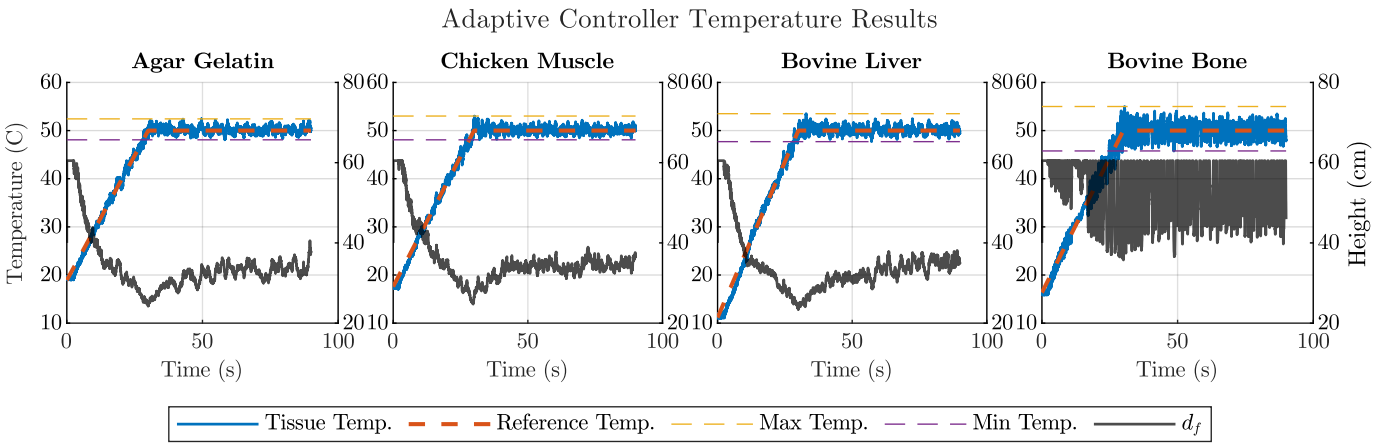

Fig. 5. Adaptive controller results. Of the five trials performed on each tissue type, here we report temperature tracking accuracy (left y-axis) and the robot height setpoint (right y-axis) where we observed the median within-group RMSE.

TABLE I
THE MEAN (STD) ROOT MEAN SQUARED ERROR (°C) OF EACH CONTROLLER'S TEMPERATURE 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)

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).

VI. Discussion

In general, our findings suggest the viability of regulating the tissue temperature at a single spot by adjusting the spot size. 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, where we observed larger RMSEs and generally noisier control signals. We note that, due to its inertia, the robot arm used in our setup certainly does not have the bandwidth to accurately follow such high-frequency control signals, which may have contributed to worse temperature tracking. In future studies, such a limitation could be overcome by adopting mechatronic laser focusing systems, such as the one described in [11], in place of a traditional robotic arm.

The degraded performance of the PI controller on bovine bone could be further explained in light of the peculiarity of bone tissue, which is significantly different in composition (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. (6)). 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 were 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 [15], 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.

Future clinical translation of the temperature control methods proposed in this manuscript has the potential to enhance robotic laser procedures. Specific targets include, among others, Fetoscopic Laser Coagulation [8], and Laser Tonsil Ablation [26]. The former is a procedure used to seal abnormal blood vessels that create a blood flow imbalance between unborn twins. The latter is an experimental surgical procedure whose purpose is to remove the tonsils, e.g., to resolve obstructive sleep apnea. In both procedures, the operating physician has to control the application of the laser to create tissue coagulation. We envision the creation of a technology that allows the physician to prescribe a certain temperature target, which would then be robotically maintained for a prescribed duration. 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).

VII. 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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