When the End Effector Is a Laser: A Review of Robotics in Laser Surgery

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Combining laser technology with robotic precision and accuracy promises to introduce significant advances in minimally invasive surgical interventions. Lasers have already become a widespread tool in numerous surgical applications. They are proposed as a replacement for traditional tools (i.e., scalpels and electrocautery devices) to minimize surgical trauma, decrease healing times, and reduce the risk of postoperative complications. Compared to other energy sources, laser energy is wavelength-dependent, allowing for preferential energy absorption in specific tissue types. This potentially leads to minimizing damage to healthy tissue and increasing surgical outcomes control and quality. Merging robotic control with laser techniques can help physicians achieve more accurate laser aiming and pave the way to automatic control of laser–tissue interactions in closed loop. Herein, a review of the state-of-the-art robotic systems for laser surgery is presented. The goals of this paper are to present recent contributions in advanced intelligent systems for robot-assisted laser surgery, provide readers with a better understanding of laser optics and the physics of laser–tissue interactions, discuss clinical applications of lasers in surgery, and provide guidance for future systems design.

1. Introduction

Robotic surgery is entering an exciting new era where robots are no longer simply mechanical instruments under the direct control of a surgeon, but rather fully-fledged cyber-physical systems able to collaborate with doctors to carry out complex procedures.[1,2] By integrating recent advances in artificial intelligence, sensing, actuation, and control, surgical robots promise to amplify the capabilities of human surgeons, enabling unprecedented levels of surgical access, precision, and safety.

Laser surgery is one of the application areas where the synergy between surgeons and the next generation of surgical robots has the potential to generate considerable clinical benefits. The history of lasers in surgery is as old as the history of the laser itself, with initial applications dating back to the early 1960s.[3] Lasers present unique capabilities that are unmatched by traditional surgical instrumentation (e.g., scalpels and electrocautery devices); in particular, a surgical laser can 1) be focused in a tiny spot (ideally, in the order of microns) and thus enable physicians to create incisions with extremely high precision; 2) cut and simultaneously coagulate tissue, thereby enabling nearly blood-less procedures; and 3) selectively target-specific tissue structures while leaving surrounding anatomy untouched (assuming that the appropriate emission wavelength is used). To date, the most successful applications of lasers are in microsurgery, which focuses on the treatment of minuscule and delicate organs, such as the ears, the eyes, and the vocal folds.

Combining lasers with robotic control can help physicians achieve more accurate laser aiming. Current state-of-the-art surgical lasers use hand-aimed mirrors, not unlike the early surgical laser systems developed nearly 50 years ago.[4] These systems require exceptional dexterity on the part of the physicians, and recent studies have shown their ergonomics to be far from ideal.[5] Furthermore, current aiming systems use traditional mechanical components that are not straightforward to miniaturize, i.e., joints and linkages. This hinders the use of lasers in many minimally invasive surgical procedures, particularly those that require the deployment of surgical instruments deep within a patient’s body. Recent advances in the fabrication of micro-electromechanical systems, such as those reported in Refs. [6,7], promise to overcome these limitations and to expand the use of lasers to new surgical applications.

There are also many untapped clinical opportunities in the control of the effects of the laser on the tissue, which can be challenging to perceive—let alone regulate—without technological aids. Lasers work in a fundamentally different manner than other clinical instruments surgeons are traditionally trained to use—e.g., scalpels. Whereas the latter uses mechanical force to cut tissue, a laser beam delivers energy, which is absorbed into the...
tissue under the form of heat, setting off a sequence of physical reactions that ultimately cause tissue vaporization.\textsuperscript{[8]} If used improperly, lasers can create thermal injuries (i.e., deep burns to the tissue and hypertrophic scarring).\textsuperscript{[9]} and no technology currently exists to help surgeons prevent or avoid these injuries.\textsuperscript{[9]} To tackle these technological challenges, several groups are investigating methods to automatically control surgical laser–tissue interactions.\textsuperscript{[10,11]} with visual or thermal sensing to close the loop. While these methods have shown promise in early proof-of-principle studies, their clinical translation presents many challenges—not least, the inherent inhomogeneity of live human tissue,\textsuperscript{[12]} which can create significant variability in the thermal response to laser pulses, and thus potentially compromise a controller’s performance.

1.1. Scope of This Manuscript

With this paper, we wish to provide a comprehensive review of the research conducted so far on robotic technology and systems for laser surgery. The two main questions that guide our review are 1) What are the opportunities that could be enabled by the combination of lasers and surgical robots? and 2) What are the technical challenges associated with the use of lasers as surgical robotic end effectors?

The contributions of this paper are as follows: 1) We provide a systematic survey of the existing body of literature on robotic systems for laser surgery and their clinical applications. Our survey identifies two main research themes, as illustrated in Figure 1. These are Robotic Actuation for Laser Aiming and Control of Laser–Tissue Interaction. For each of these two themes, we provide a comprehensive review of published work and discuss open research questions, for the benefit of current and future investigators who wish to contribute to the field. 2) We offer a brief, practical introduction to laser optics and the physics of laser–tissue interactions. In our survey of peer-reviewed papers describing surgical robotic laser systems, we found that these topics are frequently not discussed or only mentioned in passing. Yet, a working knowledge of laser optics and laser–tissue interactions is key to understanding the peculiar opportunities and challenges associated with the use of laser beams as surgical end effectors.

1.2. Manuscript Outline

The outline of this paper is as follows: We begin with a general overview of laser optics and provide a brief primer on the physics of laser–tissue interactions (Section 2), with the overarching goal of providing the necessary context to understand how surgical lasers work and how robotic assistance can help. We also discuss the current clinical applications of lasers and associated specifications. We then provide a systematic analysis of the state of the art in robotic systems for laser surgery. As it was mentioned earlier, we identify two main contribution areas, which are presented in Section 3 and 4, respectively. For each of these areas, we review recent work in the field and discuss opportunities and open challenges. Finally, Section 5 provides an overview of possible future trends and concludes the paper.

2. Background: Laser Technology and Applications to Surgery

The specifications for a surgical robot are typically based on the type of tasks that the robot is expected to perform or assist with. For instance, a robot for soft tissue surgery will generally have to be able to generate sufficient forces and torques at the end effector to manipulate tissue, e.g., for the purpose of suturing or performing incisions. Classical robot manipulation theory\textsuperscript{[13]} provides tools to aid the design of robots that meet prescribed levels of dexterity, precision, and stiffness (or compliance).

When the end effector of the surgical robot is a laser, different design considerations need to be made, which require a fundamental understanding of how laser light is generated, delivered, and absorbed into human tissue. In the following sections, we provide a brief review of the characteristics of laser light and describe the structure and main operational parameters of modern surgical laser systems. We also provide an overview of the physics of laser–tissue interaction and illustrate the effects that laser light can produce in human tissue. These topics will provide the foundation for the subsequent sections of this paper, where we will review solutions to robotically aim and control laser beams to perform surgery.

2.1. Fundamental Properties of Laser Light

Laser light is the product of stimulated emission (LASER is, in fact, an acronym for light amplification by stimulated emission of radiation. Stimulated emission was originally theorized by Einstein in his Theory of Radiation in 1917\textsuperscript{[14]}, a physical phenomenon that allows the generation of light with a high degree of monochromaticity, intensity, and directionality. A detailed review of the physics of stimulated emission is beyond...
the scope of this paper, and interested readers are referred to Ref. [15].

Monochromaticity refers to the fact that laser light is (nearly) single wavelength. The wavelength is an important parameter that determines the suitability of a surgical laser system for specific procedures. As an example, carbon dioxide (CO2) lasers are frequently used in soft tissue surgery (see, e.g., Ref. [16]), as their emission wavelength (10 600 nm) is known to be strongly absorbed by water, which is present in soft tissue in large quantities. Another common type of surgical laser is the KTP (potassium titanyl phosphate), which emits green light at a wavelength of 532 nm. The KTP laser is ideal in the treatment of highly vascularized tumors, i.e., tumors with a large blood content, as green light is strongly absorbed by hemoglobin.[17] The physical mechanisms of light absorption in biological tissue will be reviewed in greater detail in the next subsection, while a more comprehensive survey of different laser systems and their clinical applications will be provided in Section 2.4.

Intensity is defined as the amount of radiated optical power per unit of area. Laser light can be several orders of magnitude more intense than that of other light sources, which is owed to the fact that laser emitters can concentrate optical power within extremely narrow beams. Most lasers produce a circular beam whose intensity profile is described by a Gaussian function of the form[15]

\[ I(r) = I_0 \exp \left( -\frac{2r^2}{w^2} \right) \]  

where \( r \) indicates the radial distance from the center of the beam, \( I_0 \) is the peak intensity observed at the center of the beam (i.e., for \( r = 0 \)), and \( w \) is the beam radius, defined as the distance from the center of the beam where the intensity drops to \( 1/e^2 \approx 3.15\% \) of the peak value.[15] Throughout this paper, we assume that the intensity profile of any surgical laser beam can be described with Equation (1), provided appropriate values for \( I_0 \) and \( w \).

In surgical laser systems, the beam intensity cannot usually be controlled directly, and it is instead regulated by simultaneously 1) setting the optical emission power \( P \) of the system and 2) controlling the beam radius \( w \) through the use of focusing optics. It is possible to derive an equation that relates the peak intensity \( I_0 \), the emission power \( P \), and the beam radius \( w \). To do so, let us first note that \( P \) is equal to the area integral of the intensity profile \( I(r) \), which in polar coordinates is expressed as

\[ P = \int_0^{\infty} I(r) 2\pi r dr \]

\[ = \int_0^{\infty} I_0 \exp \left( -\frac{2r^2}{w^2} \right) 2\pi r dr \]

\[ = \frac{2}{\pi} I_0 w^2 \]

from which it simply follows that

\[ I_0 = \frac{2P}{\pi w^2} \]  

The relation above allows us to fully describe the intensity profile of a laser beam, given its radius \( w \) and the optical emission power \( P \). For instance, a surgical laser emitting with power \( P = 1 \text{ W} \), focused in a spot of radius \( w = 0.25 \text{ mm} \), will have a peak intensity \( I_0 = 2P/(\pi w^2) \approx 10 \text{ MW m}^{-2} \). We will come back to this relation in the next subsection, where we will use it in our discussion of how laser light is absorbed in biological tissue.

Directionality indicates a laser beam’s ability to propagate with minimal divergence. Figure 2 illustrates the propagation of a Gaussian beam in air, along an optical axis \( z \).

From Ref. [15] the radius \( w \) of a Gaussian beam varies according to the following law

\[ w(z) = w_0 \sqrt{1 + \left( \frac{z}{\alpha w_0^2} \right)^2} \]  

where \( \lambda \) is the laser wavelength and \( w_0 \) is the minimum beam radius observed at the beam waist (refer to Figure 2). The divergence of a laser beam can be more intuitively represented with a divergence angle \( \theta \), as illustrated in Figure 2, which is defined as

\[ \theta = \lim_{z \to \infty} \arctan \left( \frac{w(z)}{z} \right) \]

\[ = \arctan \left( \frac{\lambda}{\pi w_0} \right) \]  

As an illustrative example, a KTP laser (\( \lambda = 532 \text{ nm} \)), focused in a minimum radius \( w_0 \) of 0.25 mm, will present a divergence angle equal to equal to \( \theta \approx 0.7 \text{ mrad} \) (0.04°). We note that virtually all lasers used in surgical practice present a minimum beam radius of several orders of magnitude larger than the emission wavelength, i.e., \( w_0 \gg \lambda \). For these systems, therefore, the relation above can be further simplified using a small angle approximation

\[ \theta \approx \arctan \left( \frac{\lambda}{\pi w_0} \right) \approx \frac{\lambda}{\pi w_0} \]  

2.2. Thermal Response of Laser-Irradiated Tissue

Surgical lasers are designed to produce sufficient light intensity to penetrate human tissue and cause it to heat, creating the range of effects illustrated in Figure 3.

The ablation crater visible in the left image is the result of a vaporization process, where the heat produced by the laser has caused the rapid evaporation of the water present in the tissue.
As an example, the coefficient of absorption of light by tissue can be modeled using Beer’s law.

\[ I(r, z) = I_0 \exp\left(-\frac{2r^2}{w^2} - \mu_a z \right) \]  

(7)

where \( I_0 \) and \( w \) are as determined from Equation (3) and (4). In the equation above, \( \mu_a \) is called the coefficient of absorption, and it is a parameter that describes how strongly the light is absorbed by the tissue. The magnitude of this coefficient depends on the specific combination of the laser emission wavelength \( \lambda \) and the type of tissue being operated, and tabulated values are reported in the literature.\[12\] As an example, the coefficient of absorption of the CO2 laser into solid bone is estimated to be approximately 2500 cm\(^{-1}\).\[19\] In a different study, the absorption coefficient of \( \lambda = 1,110 \) nm (near infrared) laser light in human tonsils was determined to be \( \approx 0.6 \) cm\(^{-1}\).\[20\]

As the absorption coefficient has units of inverse length, its physical interpretation may not be immediately straightforward. A more intuitive parameter is the absorption length, which is simply defined as the inverse of the absorption coefficient, i.e., \( L = 1/\mu_a \). The absorption length indicates the depth along \( z \) where the light intensity drops to \( 1/e \approx 36.7\% \) of its incident value. This parameter provides a useful measure of how deeply light of a given wavelength can penetrate into a tissue target before being absorbed. With reference to the numeric examples reported earlier, the absorption length of the CO2 laser into solid bone is \( \approx 4 \) \( \mu \)m. By contrast, the \( \lambda = 1110 \) nm laser penetrates much deeper into tonsillar tissue, with \( L \approx 1.6 \) cm.

Having formulated a model to describe the laser penetration into the tissue, the corresponding temperature increase can be modeled by a function \( T(r, z, t) \), with \( t \) representing time, which satisfies the following partial differential equation:\[8\]

\[ \frac{\partial T}{\partial t} = \alpha \left( \frac{\partial^2 T}{\partial r^2} + \frac{1}{r} \frac{\partial T}{\partial r} + \frac{\partial^2 T}{\partial z^2} \right) \]
The beam radius $w$ can be controlled during a procedure by adjusting the working distance $d_f$ of the laser to the surgical site (see Figure 5). As we saw earlier, controlling the beam radius enables a physician to regulate the power density applied to the tissue. From Equation (7), tight radii enable deeper penetration and higher power density into the tissue. This is useful in the creation of laser incisions, as it ensures cutting efficiency. Conversely, larger radii will result in an overall lower power density and larger thermal spread. A physician may occasionally de-focus the laser on purpose to perform tissue coagulation, e.g., to seal a blood vessel.

To enable the treatment of anatomy seated deep within the body, which may be difficult to reach in a line-of-sight fashion, some surgical laser systems deliver light by means of optical fibers. In these systems, the beam radius is still controlled by regulating $d_f$, which in this case indicates the distance between the fiber tip and the tissue surface. It should be noted that light exiting an optical fiber typically diverges with an angle determined by the numerical aperture of the fiber.[21]

### 2.3.2. Beam Modulation

Surgical lasers can emit light either as a continuous wave or in pulsed mode. When operated in continuous wave (CW) mode, a laser will simply produce a constant optical power output $P$. In pulsed mode, the output power is modulated to create a train of pulses at the desired repetition rate $f$ with a prescribed pulse width $\tau$. The shape of individual pulses can be modulated to achieve different effects on the tissue. Clinically, the pulsed mode is preferred, as it has been shown to create less heat spread, therefore posing a lower risk of thermal injury.[22]

### 2.4. Clinical Applications of Lasers

Lasers are used in various surgical applications due to their preciseness and effectiveness in ablating minuscule-sized and delicate tissues. The quality of laser ablation depends on several factors including the focus of the laser, laser sources, power settings, and exposure time.[18] By controlling the focus of the laser, the ablation size can be varied from macroscale to microscale.[23] In free beam-based systems, focusing optics are embedded to precisely control the spot size of the laser beam. Whereas fiber steering-based systems do not include such optical lenses, and the laser beam diverges as soon as it exits from the optical fiber. Thus, the spot size of the laser beam needs to be adjusted manually by controlling the distance between the laser and the target tissue. For this reason, the fiber steering-based systems do not have a fixed range of spot size, as shown in Table 1.

![Figure 5. Structure and main operational parameters of surgical laser systems. (Left) A laser emitter produces a beam of light with optical power $P$. The power output can be modulated as a train of pulses with duration $\tau$ and rate of repetition $f$. The beam diameter $w$ is regulated by controlling the focal distance $d_f$ between a focusing lens and the tissue surface. (Right) To avoid the line-of-sight limitations of traditional free beam lasers, and thus enable access to anatomy seated deep within the body, some laser systems deliver light by means of optical fibers. The radius of the laser beam can still be controlled by regulating $d_f$, which in this case indicates the distance between the fiber tip and the tissue surface. Light exiting an optical fiber typically diverges with an angle determined by the numerical aperture of the fiber.[21].](Image)
Currently, laser-assisted surgeries are performed in fetal, urology, laryngology, ophthalmology, and otology surgical procedures. In these surgeries, various types of tissues from soft to hard tissues are ablated, and these tissues have different coefficients of absorption.\(^{[18]}\) To leverage the coefficient of absorption, laser sources and power settings are varied depending on the target tissues, as listed in Table 1. With regard to laser delivery, lasers are manually guided and steered by the surgeon either using an optical fiber (i.e., fiber steering) or optical lenses (i.e., free beam), as listed in Table 1 and discussed in the following. Examples of laser-assisted surgeries with a fiber steering approach are described below.

In fetal surgeries, an optical laser fiber and a fetoscope are inserted into a rigid sheath with an \textit{Albarran} deflector mechanism to move through a tortuous path. Then, as illustrated in Figure 6, the placental vessels are coagulated using the laser.\(^{[24–27]}\) To effectively ablate tissues during fetal surgeries, laser sources like Nd:YAG and 940 nm diode laser are utilized because their energy is strongly absorbed by hemoglobin.\(^{[24]}\)

**Table 1.** Laser specifications of laser-assisted surgeries.

| Surgery \(^{[a]}\) | Required Accuracy [μm] | Laser Source | Laser Steering Method | Laser Wavelength [nm] | Power Range [W] | Fiber Core Diameter [μm] | Fiber Material | Spot Size [μm] |
|--------------------|------------------------|--------------|-----------------------|-----------------------|-----------------|--------------------------|----------------|----------------|
| Fetal surgery\(^{[24,113–117]}\) | 250 | Nd:YAG \(^{[b]}\) | FS\(^{[c]}\) | 1064 | 50–100 | 400–600\(^{[d]}\) | Silica | Variable |
| – | – | Diode | FS\(^{[c]}\) | 940 | 20–60 | 400–600\(^{[d]}\) | Silica | Variable |
| Urology\(^{[23,28,115–122]}\) | – | Ho:YAG | FS\(^{[c]}\) | 2100 | 60–120 | 450–1450 | Silica | Variable |
| – | – | Thulium fiber | FS\(^{[c]}\) | 1920–1960 | 2–60 | 315–1500 | Silica | Variable |
| – | – | Nd:YAG | FS\(^{[c]}\) | 1060 | 60–100\(^{[e]}\) | 600\(^{[f]}\) | Silica | Variable |
| – | – | Tm:YAG | FS\(^{[c]}\) | 2000 | – | – | Silica | Variable |
| – | – | KTP | FS\(^{[c]}\) | 532 | 0–80 | – | Silica | Variable |
| Laryngology\(^{[115,117,118,123]}\) | 50 | CO\(_2\) | FS\(^{[c]}\) | 1060 | 1–40 | – | Silica | Variable |
| – | – | FB\(^{[c]}\) | 1060 | 1–40 | – | – | 0.4–4.0 |
| Ophthalmology\(^{[40,41,115,117,124]}\) | 100 | OPSL\(^{[d]}\) | FS\(^{[c]}\) | 532, 577 | 0–2 | 200–600 | Silica | Variable |
| – | – | FB\(^{[c]}\) | 532, 577 | 0–2 | – | – | 0.4–4.0 |
| Otology\(^{[32,115,117,125–127]}\) | 400 | CO\(_2\) | FS\(^{[c]}\) | 1060 | 1–5\(^{[g]}\) | 100–600 | Silica | Variable |
| – | – | KTP | FS\(^{[c]}\) | 532 | 1\(^{[h]}\) | – | Silica | Variable |
| – | – | Er:YAG | FS\(^{[c]}\) | 2940 | – | 400, 450 | ZrF\(_4\)^{[i]}, GeO\(_2\)^{[g]} | Variable |

\(^{a}\) This table was generated by referring to clinical journal papers and specifications of commercially available surgical laser systems. \(^{b}\) FS: Fiber steering. \(^{c}\) FB: Free beam. \(^{d}\) OPSL: Optically pumped semiconductor laser. \(^{e}\) Clinically used laser parameters. \(^{f}\) ZrF\(_4\): Zirconium fluoride. \(^{g}\) GeO\(_2\): Germanium oxide.

**Figure 6.** Clinical applications of laser-assisted robotic surgeries and steering methods. With the fiber steering approach (top), the laser is oriented by rotating and translating an optical fiber via a surgical access port. With the free beam method (bottom), the laser beam is oriented by using mirrors.
For urology surgeries, laser-assisted surgeries are performed by inserting an optical fiber into the working channel of a resectoscope. The laser is aimed using the fiber steering approach, and the surgeon manually maneuvers the resectoscope and the laser fiber to visualize the target surgical area and perform the operation. During these procedures, the fiber and resectoscope are in a liquid environment. As illustrated in Figure 6, one of the examples of these surgeries is prostate laser enucleation. In this procedure, surgeons use laser sources (see Table 1) with a characteristic wavelength that is strongly absorbed by water or proteins to appropriately incise, coagulate, and resect soft tissues. These laser sources include 1) holmium:yttrium–aluminum–garnet (Ho:YAG), 2) neodymium:yttrium–aluminum–garnet (Nd:YAG), 3) thulium:yttrium–aluminum–garnet (Tm:YAG), and 4) potassium titanyl phosphate (KTP). In addition to ablating soft tissues, the lasers used in urological surgery are also applicable for fragmenting hard tissues like kidney stones within the urinary tract. A smaller size of optical laser fiber, varying 315 μm to 1.5 mm, needs to be changed because the size of the lumen differs within the urinary tract. A smaller size of optical fiber allows reaching deeper within the urinary tract; however, the fiber becomes more fragile at the same time.

For ear surgeries, as shown in Figure 6, laser energy is delivered using an optical fiber to eliminate cholesteatoma, a cyst that grows behind the eardrum and invades ear structures, or to create a path to insert a prosthesis. KTP, carbon dioxide lasers (CO2), and erbium-doped yttrium aluminum garnet (Er:YAG) are often used for ear surgery because both soft and hard tissues within the ear are ablated using these laser sources. The laser generated by KTP is strongly absorbed by hemoglobin and melamin in general. Whereas, the laser lights provided by CO2 and Er:YAG are highly absorbed by the water inside the bone, producing micro-explosion within the bone to ablate.

Lasers have been employed in transoral surgery using both fiber steering and free beam approach. For the fiber steering approach, the optical laser fiber is mounted on a grasping, which is operated by a surgeon to control the laser beam. During the surgery, the laser is adopted to ablate abnormalities such as cancers, polyps, nodules, and cysts in the vocal cords. As reported in Table 1, a CO2 laser is used for this type of surgery because soft tissues have a high-water content, which strongly absorbs CO2 laser light.

Similarly, both the fiber steering and free beam approaches are adopted in eye surgery. Generally, the fiber steering approach is used for surgeries performed in the operating room, where patients with advanced symptoms are treated. Examples of these surgeries include anterior eye surgeries and intraocular surgeries. During anterior eye surgeries, patients are not always anesthetized, rather they are only locally sedated. Hence, there have been challenges in delivering laser precisely and accurately not only because of the surgeons' hand tremors but also involuntary patient movements that can affect surgical outcomes. In the case of intraocular surgeries, microsurgical tools including laser and light source are entered through the pars plana (i.e., the anatomical part of the eye located near the junction of the iris and sclera) as illustrated in Figure 6. During these procedures, surgeons often encounter problems due to limited visualization, poor maneuverability, and surgeons' hand tremor. To address these challenges, surgeons would either need a robotic fixture that fittingly stabilizes the patient’s eye with a proper amount of force or a robotic laser steering device that could apply laser energy to the appropriate location by compensating for hand tremor or via real-time target tracking.

As demonstrated by these clinical applications, the benefits of the fiber steering method are the ability to deliver laser energy inside the body in a minimally invasive fashion and position the laser fiber tip in front of the surgical site, providing a short working distance. Thus, this permits the laser beam to have an unobstructed path between the laser fiber and the target tissue, securing a direct line of sight, despite the need to navigate to the target surgical area via natural orifices or small incisions on the patient. However, the current fiber steering method presents several technical challenges.

Laser surgical procedures that present the advantage of a direct line of sight (i.e., do not require minimally invasive access to the surgical target via navigation through natural orifices or incisions) are typically performed using the free beam method. Currently, the free beam steering approach is adopted in eye surgeries and vocal cord surgeries.

In the case of eye surgeries, laser photocoagulation is performed to treat various retinal diseases. These procedures require high accuracy (100 μm) in terms of the size and position of laser spots. To satisfy those requirements, slit lamp delivery systems and motorized laser photocoagulation devices, such as Navilas (OD–OS GmbH, Teltow, Germany) and PASCAL (IRIDEX, Mountain View, CA, USA), are used. These devices use the free beam-based laser delivery mechanism, a galvanometric scanner, to deliver the laser beam to predefined target spots. Generally, these semi-automated systems are designed for outpatient clinics.

Another example of a surgical procedure performed with a free beam mechanism is laser-assisted vocal cord surgery. As illustrated in Figure 6, traditional laser-assisted vocal cord surgery is performed using a laryngoscope, a surgical microscope, and a laser micromanipulator which consists of a beam splitter mirror and a mechanical joystick. This method requires surgeons to manually control the micromanipulator to fire CO2 laser beam spot-by-spot. The typical working distance of laser-assisted vocal cord surgery is about 400 mm, and the size of the operation site is ~1 mm. Due to such operating conditions, precise ablation of tissue is challenging. For these surgeries, surgeons need to accurately dissect the pathology while minimizing the thermal damage to healthy tissue. Extensive accumulation of heat to healthy tissue can result in poor clinical outcomes such as the carbonization of tissues. Hence, such operating conditions require highly accurate (50μm) and precise laser manipulation. To mitigate those problems, mirrors are motorized to provide high positional accuracy and support the laser scanning functionality. These features help to minimize undesirable thermal damage to the surrounding tissues and ensure a better quality of laser–tissue interaction during transoral surgery compared to the fiber steering systems described above. The commercially available free beam scanners include Digital AcuBlade Scanning Micromanipulator (Boston Scientific Corp., Marlborough, MA, USA) and SoftScan Plus R (KLS Martin Group, Tuttingen, Germany). These devices...
improve the quality of laser–tissue interaction by rapidly generating preprogrammed laser scan patterns and controlling the depth of ablation.\textsuperscript{[43]}

The motorized free beam approach provides high positional accuracy (75 μm) and fast laser steering speed (200 mm s\textsuperscript{-1}).\textsuperscript{[50]} However, the currently available free beam laser systems need to be mounted onto a microscope, resulting in a long working distance (400 mm). With this given condition, the surgical site can easily be obstructed by the tissues in the pathway and surgeons can lose the direct line of sight to the surgical site. To resolve these problems, the size of laser steering mechanisms needs to be miniaturized to the millimeter scale. The miniaturization of the free beam laser steering system would place the motorized mirrors closer to the surgical site. This would secure the direct line of sight to the surgical site while maintaining the speed and positional accuracy of the system.

3. Robotic Systems for Laser Surgery

In recent years, there have been attempts in combining surgical lasers with commercially available surgical robotic platforms to alleviate the aforementioned issues in laser beam steering. Mainly, these attempts have been made in performing laser-assisted transoral surgeries, where laser fibers are integrated into the end effectors of commercially available surgical robots. Examples of these include Fiberlase (Boston Scientific Corp., Marlborough, MA, USA)\textsuperscript{[51]} and Flexguide Ultra (OmniGuide Inc., Cambridge, MA, USA).\textsuperscript{[52]} These are attachable surgical laser fibers that can be mounted onto a robotic grasper. These devices are intended to be used with the da Vinci SP (Intuitive Surgical Inc., Sunnyvale, CA, USA),\textsuperscript{[53]} a surgical robot for single port surgery with a camera and three manipulators, to perform laser surgery in a small and confined space.\textsuperscript{[54]} Likewise, Flex robotic system (Robotic Surgical System, Dubai, UAE)\textsuperscript{[55]} is a flexible robotic endoscope that enables laser-assisted transoral surgery by manually steering an optical fiber. For this system, two external working channels are attached to the robotic endoscope such that they follow the movement of the endoscope. Through these channels, various surgical instruments including an optical fiber can be inserted and manually controlled by a surgeon.\textsuperscript{[56]}

The integration of laser fibers within these robotic platforms introduced several advantages, such as increased precision and safety in resecting tissues using a laser by placing the camera 100 mm apart from the target\textsuperscript{[57]} and manipulating the laser beam in front of the surgical site. In addition, the da Vinci robotic platform eliminated the hand tremor induced by the operator.\textsuperscript{[57,58]} This minimized the risk of delivering laser energy to unintended targets and reduced the undesired tissue contact, which could cause carbonized tissue and tissue sticking problems.\textsuperscript{[59]} However, the aforementioned integration did not completely address the shortcomings of manual control of surgical lasers because these robotic platforms were not specifically designed to perform laser-assisted surgeries. Hence, problems such as providing the level of accuracy required by specific types of surgery, steering the laser at the speed of conventional laser scanners, and controlling the focus of the laser beam remained.

To overcome the limitations of current laser systems and improve clinical outcomes of laser surgical procedures, many research groups have been developing dedicated robotic systems for laser surgery. These robots exploit various types of actuators, i.e., motors, magnetic actuators, fluidic actuators, and piezoelectric actuators, to control the tip of an optical fiber or the orientation of optical lenses to guide the laser beam (Figure 7). In this section, we present recent contributions in robotic systems for laser surgery according to the following design considerations: 1) we discuss their clinical applications and laser delivery mechanisms; 2) we present actuation strategies and methods for laser steering; 3) we analyze the engineering requirements and resulting performance of these robots in terms of size and speed, range of motion, optics, and laser characteristics; and 4) lastly, we address the strengths and limitations of each actuation strategy.

Figure 7. Actuation methodologies for traditional and robotic laser steering mechanisms in surgery.
3.1. Laser Fiber Steering Robots

The advancement of optical fibers in flexibility, biocompatibility, and light transmission has expanded the range of clinical applications from soft tissue to hard tissue laser surgeries.[32] However, the design of a robotic platform that could control a laser optical fiber in a precise and accurate manner remains a challenge.

To address this shortcoming, researchers have been developing robots that are specifically designed to perform laser-assisted surgeries by steering optical fibers. These research platform robots are classified depending on their actuation strategy, i.e., motor-driven, piezoelectric actuated, magnetically driven, and fluidically actuated robots.

In Figure 8, we list and compare the actuation strategy, engineering requirements, and clinical applications of research platforms. For the engineering requirements, we primarily focus on informing mechanical aspects of the robots, i.e., size, range of motion, speed, and mechanical bandwidth and describing optical components, i.e., laser focusing optics, and laser characteristics.

For minimally invasive surgeries, the size of the robot is mainly constrained by the size of the natural orifice or instrument port through which it needs to be inserted. If the size of the robot exceeds this constraint, then the robot cannot be deployed and defeats the purpose of minimally invasive surgery. The range of motion describes the area in which a focused laser beam can be delivered. Depending on the laser steering approach, this range of motion can be reported in bending angle because the laser can be steered and delivered three dimensionally through an optical fiber. Considering that each laser steering approach can provide a different range of motion and the size of ablating tissues varies for each type of surgery, an appropriate laser steering approach needs to be selected to meet the requirements. Speed, mechanical bandwidth, and positional accuracy of the robot directly relate to the results of tissue ablation. Faster and more accurate laser steering would minimize the thermal damage to healthy tissues and increase the range of clinical applications. Likewise, laser focusing optics and laser characteristics affect the efficiency of the laser ablation. Therefore, considerations of optical designs and laser characteristics are needed to expand the clinical feasibility.

3.1.1. Motor-Driven Continuum Robots

A common approach to steering surgical laser fibers with robots is by using motors. Broadly categorized, three types of motor-driven robotic manipulators have been proposed in the literature to steer optical fibers, i.e., backbone-based manipulators,[60–65] concentric tube robots,[66] and hybrid concentric tube robots.[67] For these approaches, the robots have a flexible distal end, and this flexible end controls the movement of an embedded optical fiber. Typical transmission components to control the distal end include cables[60,62,65,68] and push-pull rods,[64,66] and they are mostly fabricated and assembled from off-the-shelf components. As a result of those transmissions, the robots can provide a wide range of motion (from 2.5 to 40 mm). However, at the same time, some transmission components (i.e., cables) suffer from nonlinearity caused by friction. To compensate the nonlinearity and external disturbances, the robots...
include a controller such as visual feedback controllers,\textsuperscript{[65,68]} cable tension controllers\textsuperscript{[69]} and motor joint speed controllers\textsuperscript{[63,66]} to keep the position error below 1 mm, as shown in Table 2.

A backbone-based manipulator is a common design of continuum robots. As shown in Figure 9a, the backbone-based manipulator consists of an elastic structure with cavities for a working channel and actuation cables. Using an elastic structure such as Nitinol (NiTi) tubes\textsuperscript{[63]} or polymer,\textsuperscript{[62,64,65]} the manipulator is capable of bending upon actuation and restoring its original position once the actuation force is released. Using soft materials capable of bending upon actuation and restoring its original position once the actuation force is released, as shown in Table 2, these manipulators can be miniaturized down to 2.1 mm (excluding a camera and an illumination channel), and they provide a large range of motion between (30 mm × 30 mm)– (45 mm × 45 mm) by implementing appropriate transmission

| Author                  | Steering Method | Actuator                  | Size [mm] | Range of Motion [mm] | Maximum LaserBending Angle [°] | Maximum Speed [mm/s] | Mechanical Bandwidth [Hz] | Positional Accuracy [mm] | Laser Focusing | Laser Source | Clinical Application          |
|------------------------|-----------------|---------------------------|-----------|----------------------|-------------------------------|----------------------|---------------------------|---------------------------|----------------|-------------|-----------------------------|
| Harada et al.\textsuperscript{[60]} | Fiber           | Motor w/ cable            | 2.4       | –                    | 90                            | –                    | –                         | –                         | X              | Nd:YAG      | Fetal surgery               |
| Yamashita et al.\textsuperscript{[81]} | Fiber           | Motor w/ linkages         | 3.5       | 25 × 50\textsuperscript{[1]} | 80                            | –                    | –                         | 0.2                       | X              | Nd:YAG      | Fetal surgery               |
| Goldman et al.\textsuperscript{[63]} | Fiber           | Motor                     | 5         | φ 40                 | –                             | –                    | –                         | –                         | X              | Holmium fiber laser         | Urology surgery          |
| Russo et al.\textsuperscript{[62]} | Fiber           | Motor w/ cable            | 2.1       | φ 6                  | –                             | –                    | –                         | –                         | X              | Ho:YAG      | Urology surgery             |
| Hendrick et al.\textsuperscript{[86]} | Fiber           | Motor                     | 2         | –                    | –                             | –                    | –                         | –                         | X              | Ho:YAG      | Urology surgery             |
| Ferhanoglu et al.\textsuperscript{[6]} | Fiber           | Piezoelectric tube        | 5         | 0.25 × 0.25          | –                             | 450\textsuperscript{[1]} | 895                       | –                         | X              | Erbium fiber laser          | –                        |
| Yang et al.\textsuperscript{[36,112]} | Fiber           | Piezoelectric motor       | –         | φ 4                  | –                             | 5                    | –                         | 0.05                      | X              | 532 nm laser              | Ophthalmology Surgery    |
| Acemoglu et al.\textsuperscript{[54,55,73,72,28]} | Fiber           | Electromagnet             | 13        | 4 × 4                | –                             | 94                   | 63                        | 0.09                      | O              | 625 nm laser              | Laryngology Surgery      |
| Chikhaoui et al.\textsuperscript{[88]} | Fiber           | Electroactive Polymer     | 2         | φ 30\textsuperscript{[4]} | 5                             | 0.166\textsuperscript{[2]} | 0.05\textsuperscript{[2]} | 0.12                      | X              | –                        | –                        |
| Charreyron et al.\textsuperscript{[74,120]} | Fiber           | Ferromagnetic             | 0.85      | 26\textsuperscript{[9]} | 90                            | –                    | –                         | <0.16                     | X              | 780 nm laser              | Ophthalmology Surgery    |
| Kim et al.\textsuperscript{[76]} | Fiber           | Ferromagnetic             | 0.5       | –                    | –                             | –                    | –                         | –                         | X              | –                        | –                        |
| Kundrat et al.\textsuperscript{[84,110]} | Fiber           | Motor w/ push-pull rods   | 11.5      | 45 × 45              | <180                          | 3.5                   | –                         | 0.25                      | O              | 650 nm laser              | Laryngology surgery      |
| Zhao et al.\textsuperscript{[89]} | Fiber           | Motor w/ cable            | 22        | 28\textsuperscript{[9]} | –                             | 3                    | –                         | 0.054                     | X              | Thulium Laser            | Laryngology surgery      |
| Fang et al.\textsuperscript{[91]} | Fiber           | Hydraulic                 | 12        | 15 × 15              | 26                            | 2                    | 1                         | <0.2                      | X              | 808 nm laser              | Laryngology surgery      |
| Li et al.\textsuperscript{[85]} | Fiber           | Motor w/ cable            | 9         | 59.4                 | –                             | –                    | –                         | 1.4                       | X              | 690 nm laser              | General Surgery           |
| Mo et al.\textsuperscript{[68]} | Fiber           | Motor w/ cable            | 3         | –                    | –                             | –                    | –                         | –                         | X              | –                        | –                        |
| Nguyen et al.\textsuperscript{[67]} | Fiber           | Motor w/ cable            | 2.5       | –                    | 27                            | –                    | –                         | 0.687                     | X              | 532 nm laser              | Otolaryngology Surgery    |
| Yamanaka et al.\textsuperscript{[80]} | Free beam       | Motor                     | 7         | 10.7 × 9.7           | 24                            | –                    | –                         | <1                        | O              | Nd:YAG      | Fetal surgery               |
| Patel et al.\textsuperscript{[84]} | Free beam       | Risley prism w/ piezoelectric motor | 17 | –               | –                             | –                    | –                         | –                         | –              | O                        | CO₂ Laser                 |
| Rabenrososa et al.\textsuperscript{[83]} | Free beam       | Piezoelectric             | 8 × 10    | >20 × 20             | 45                            | 33\textsuperscript{[4]} | 273                       | 0.1                       | X              | –                        | Laryngology surgery      |
| Bothner et al.\textsuperscript{[87]} | Free beam       | Piezoelectric             | 15        | 18 × 10              | –                             | 2,000                | 750                       | –                         | O              | –                        | Laryngology surgery      |
| Renevier et al.\textsuperscript{[84]} | Free beam       | Piezoelectric             | 9 × 11    | 20 × 20              | 45                            | 33\textsuperscript{[4]} | 273                       | 0.0805                    | X              | Er:YAG       | Laryngology surgery         |
| York et al.\textsuperscript{[7]} | Free beam       | Piezoelectric             | 6         | 18 × 18              | –                             | 3,900                 | 1,200                     | –                         | O              | –                        | Laryngology surgery      |

\textsuperscript{[1]}Estimated value based on published data.
components. As demonstrated by the performance of these robots, the range of motion of motor-driven robots is an order of magnitude greater than that of the robots driven using other types of actuators, i.e., piezoelectric, magnetic, electromagnetic, and hydraulic. This large range of motion helps to avoid frequent re-positioning of robots if a surgeon needs to ablate malignant tissues within a large area, and this could reduce the operation time of laser-assisted surgeries as well.

A concentric tube manipulator is another manipulator design of a continuum robot. A concentric tube manipulator nests a series of preshaped, elastic tubes made of NiTi, as shown in Figure 9b. To enable the motion of the concentric tube manipulators, the nested tubes are translated and rotated within each other by using multiple motors. Inside the tubes, various surgical instruments including an optical fiber can be passed through and manipulated simultaneously. Using this robotic
innovation, Hendrick et al.\cite{66} applied this manipulator to perform laser-assisted benign prostatic hyperplasia (BPH) treatment. In their design, two concentric tube robots were passed through the 5 mm working channel of a resectoscope. In contrast to the traditional approach of laser-assisted BPH surgery, which entails using only one surgical instrument (i.e., a laser fiber) in the resectoscope working channel, this robot was capable of adding an extra manipulator with multiple degrees of freedom. As a result, this robotic approach increased the surgeon’s dexterity and enhanced tissue manipulation capabilities. In addition, the reachable area using the robotic platform increased by 65% with respect to the standard (i.e., manual fiber steering) endoscopic method.

Combining the concept of a backbone-based robot and a concentric tube robot, a hybrid concentric tube robot has been proposed (see Figure 9c). Overall, the robot actuates similarly to a concentric tube robot, with nesting concentric tubes to steer the manipulator. However, instead of only including flexible tubes made of NiTi, various other tube designs, such as a fixed curvature tube and a cable-actuated backbone-based tube, are included to improve the workspace and force transmission of the robots. In the work of Nguyen et al.\cite{67} the researchers designed a hybrid concentric tube robot to allow surgeons to navigate and perform laser surgery within a confined, intricate space during cholesteatoma laser surgery. The proposed robot consists of three tubes, a straight tube made of stainless steel, a precurved tube made of NiTi, and a backbone-based tube that is capable of being actuated and made of NiTi. Using this series of tubes, they enhanced the dexterity of the robot and further increased the coverage volume of the robot (434 mm$^3$), which is about twice as large as that by solely using the backbone-based robot or concentric tube robot.

Although motor-driven systems have many strengths in comparison to the traditional fiber steering-based surgical procedures, there are remaining challenges. First, the laser steering speed of these robots is relatively slow in comparison to other actuation strategies, as illustrated in Figure 10. The maximum laser steering speed of these robots is below 100 mm s$^{-1}$, making them unsuitable for certain laser surgical applications. For example, the laser scanners that are used for vocal surgeries provide 200 mm s$^{-1}$ laser scanning speed to ablate tissues. Such speed ensures the quality of laser ablation by minimizing thermal damage to the surrounding tissue. Second, the motor-driven systems generally suffer from nonlinearity induced by friction. To address this problem, many of the proposed robots are teleoperated, leaving surgeons to mitigate the problem through visual feedback. Nevertheless, the ultimate goal of these surgical robots would be increasing the autonomy to the point where a human operator is not needed.\cite{2} To do so, intricate control algorithms or the use of different mechanisms such as cable-driven parallel mechanism and linkage-based mechanisms\cite{61,69} will need to be further investigated to compensate the nonlinearity that could be caused by friction and hysteresis.

3.1.2. Piezoelectric Actuator Driven Robots

One of the inherent problems of the fiber steering approach is slow laser steering speed that hampers the laser scanning functionality. The laser scanning functionality is an important feature in various surgeries because it helps to reduce heat concentration by rapidly moving the laser beam and prevent tissues from carbonization. Motivated by such a limitation, piezoelectric actuators have been proposed to increase the laser fiber steering speed and the mechanical bandwidth. Under the stimulus of high voltage (typically exceeding 100 V), a piezoelectric actuator converts the electrical energy into mechanical displacement and provides fast response with μm-scale positional accuracy.\cite{71} In addition, the variety of form factors (i.e., tube, plate, and disc) and their millimeter-scale packaging sizes make piezoelectric actuators an attractive choice of an actuator to embed within a laser steering mechanism.

Using the tube type piezoelectric actuator, Ferhanoglu et al.\cite{6} developed a laser steering robot. They attached a piezoelectric tube actuator directly to an optical fiber and forced the fiber to

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**Figure 10.** Comparison of different actuation strategies for laser surgery robots: laser steering speed vs size (i.e., robot diameter).
physically move to steer the laser beam. As shown in Table 2, they achieved fast laser ablation speed (270 mm s$^{-1}$), thus allowing rapid ablation of the target surface with the laser beam. Likewise, Yang et al.\cite{36} developed a fiber steering robot for intraocular laser surgery. As illustrated in Figure 9d, the design of their robot includes a Gough–Stewart platform and six piezoelectric linear actuators to control the movement of the attached optical fiber. An optical tracking system tracks the position and orientation of the optical fiber to effectively control its position and compensates motion disturbance caused by hand tremor. The robot has 6 degrees of freedom, provides 4 × 4 mm range of motion, and accurately guides the laser to the predefined target using visual feedback. Despite these successful examples of piezoelectric actuator integration, there are several drawbacks that need to be considered for applications in robotic laser surgery.

One of the inherent characteristics of piezoelectric actuators is that they provide small displacements (i.e., 10 μm–6 mm depending on the type of piezoelectric actuators). Such limited stroke results in a reduced range of motion while steering the laser fiber as shown in Table 2. Therefore, depending on the type of surgery, piezoelectric actuator-driven robots may require frequent re-positoning to ablate an entire block of malignant tissue. Lastly, piezoelectric actuators generally require voltages in the order of 100 V to actuate. This can raise safety concerns, especially if the surgical instrument is used inside the human body.

3.1.3. Magnetic Actuator Driven Robots

Another emerging approach for laser fiber steering is magnetic actuation. Magnetic actuation generates the necessary force and torque to operate a robot through the interaction of magnetic fields, where the common sources of magnetic field include permanent magnets and electromagnets. Based on its governing principle, magnetic actuation needs at least two magnetic field generators to initiate the interaction, where one of the sources can either be located inside or outside the robot.\cite{72–76} Using this actuation strategy, it has been possible to further miniaturize the size of robots for laser surgery, remotely control the robot depending on the configuration, and achieve relatively fast actuation within a confined space.

Adopting the aforementioned advantages, there has been an attempt to integrate magnetic actuation to rapidly steer an optical fiber and improve tissue ablation quality. In the work done by Acemoglu et al.\cite{44,79} the authors embedded the magnetic actuation system to steer the tip of an optical fiber in multiple directions for vocal cord surgery. In their actuator design, as illustrated in Figure 9e, a ring permanent magnet was attached to an optical fiber and embedded four sets of miniature electromagnetic coils around the permanent magnet. Although placing the electromagnetic coils within the robot enlarged its overall diameter to around 10 mm and limited the range of optical fiber motion within the enclosure of the robot, the optical fiber could be maneuvered with low power (under 1.5 W). In addition, two plano-convex lenses were attached at the tip of the robot to provide a collimated and focused laser beam. Using the mechanism, they developed a system with laser speed of 94 mm s$^{-1}$ and a mechanical response of 63 Hz. The range of motion (4 × 4 mm) was larger than that of the piezoelectrically actuated robots.

To develop smaller scale fiber steering robots, several groups adopted off-board magnetic actuation strategies\cite{74,76,77} which use an external magnetic field to control a robot. For this method, the main body of the robot includes magnetic sources like permanent magnetic or ferromagnetic particles, and the robots are controlled using the external magnetic field generated by permanent magnets and electromagnets. In the work of Gervasoni et al.\cite{77} and Chareyron et al.\cite{74} they implemented off-board electromagnetic actuation to develop robots that aimed at performing laser-assisted fetal surgery and ophthalmologic surgery. Both surgeries require navigation within confined spaces, i.e., less than 4 mm cavity for fetal surgery and 1 mm cavity for ophthalmologic surgery. Thus, both robots used optical fibers to deliver the laser beam, and the electromagnets that induce the motion of the robots were controlled from outside the patient to meet the strict size requirements. As a result, the size of the proposed robots was scaled down to fit within an 11 Fr trocar and 850 μm, respectively. In the recent work from Kim et al.\cite{76} they demonstrated the possibility to further miniaturize magnetically actuated fiber steering robots for endovascular neurosurgery applications (Figure 9f). The main structure of their ferromagnetic soft continuum robot was fabricated with a soft, ferromagnetic polymer. Silica coating and a hydrogel skin were added to prevent corrosion of the ferromagnetic microparticles and to induce smooth motion of the robot by reducing the surface friction between contacting tissue and the robot. The overall size of the robot is 500 μm, and it demonstrated successful navigation through a phantom 3D cerebrovascular network and a phantom carotid artery.

Despite the strengths demonstrated by magnetically actuated laser surgery robots, there are shortcomings associated with this actuation strategy. To start with, the use of electromagnetic coils introduces heating problems. This could be a potential hazard if the robot embeds onboard electromagnetic coils to steer the laser beam. As the electromagnets generate magnetic fields using current, the electromagnets would produce heat within the robot. Thus, effective dissipation of heat without impacting the operating environment remains an unmet challenge. Moreover, the current approach to miniaturization of magnetic actuator-driven robots is typically accomplished by using off-board magnetic actuation (i.e., placing the controlling magnetic field generator outside the robot).\cite{72} However, this comes at the expense of adopting a larger-scale, stronger magnetic field generator because the distance between the robot and the magnetic field generator increases. Such a problem can possibly lead to disruption of the functionality of other medical instruments that are within the human body or in the operating room, and metallic surgical tools (i.e., scalpels, forceps).\cite{78}

Although those problems can be mitigated by placing the magnetic field generator closer to the robot to minimize the distance between the two and reduce the required magnetic field, this may not be an option depending on the clinical application. Thus, the future implementation of magnetic actuation for optical fiber steering should further consider the trade-offs between the size of the robot and its performance (i.e., range of motion and required power).

3.1.4. Fluidic Actuator-Driven Robot

With recent advancements in soft surgical robotics, fluidic actuators have been utilized as a new type of actuation mechanism for...
surgical laser steering.[79] Unlike other types of actuators, fluidic actuators do not rely on high electric power or magnetic field to operate.[80] This not only ensures the safety of both patients and surgeons but also guarantees compatibility with intraoperative imaging devices (e.g., magnetic resonance imaging (MRI)), which can be used to monitor the depth of laser ablation.[9] For piezoelectrically or electromagnetically driven robots, this would not be possible because the actuators would induce artifacts on images.[81] Moreover, soft robots allow safe interaction with human biological tissues due to their inherent flexibility and compliance.[82] Nevertheless, these soft robots possess several drawbacks including size, noise, leakage from fluidic actuators, slow actuation speed, and inaccuracy in motion control.[78] The viscoelastic properties of soft polymer materials introduce hysteresis and nonlinear behavior in open loop conditions. The slow response caused by fluidic actuation further complicates the control of nonlinear behavior and hinders the movement of optical fibers.[83] The accuracy and speed of robot motion are important aspects of surgical laser robots because inaccuracies in position and speed can lower ablation quality by damaging healthy tissues.

In the recent work by Fang et al.,[81] they proposed a soft robot to steer an optical fiber using soft chambers and a hydraulic actuator and addressed inherent drawbacks of soft robots with a neural network-based controller (Figure 9g). The three hyperelastic chambers were inflated with water to effectively steer an optical fiber that is located at the center of three chambers. Using this actuation strategy, the robot was able to steer the laser within a 15 × 15 mm area, which is sufficient range of motion for transoral surgery. However, as a result of using hyperelastic materials for soft chambers, inherent nonlinearity and hysteresis of soft robots were introduced. To address this problem, the robot included spring reinforcement constraints to stiffen the chambers and reduce hysteresis. Additionally, the robot used a neural network-based controller to minimize the effect of the nonlinearity. As a result, the positioning error was lowered to less than 0.2 mm. In spite of these improvements, the slow laser steering speed, and low bandwidth of the robot caused by the use of the hydraulic actuator remained challenge. The laser steering speed of the robot reported in Ref. [81] was in the order of 1 mm s\(^{-1}\). As shown in Figure 10, the soft hydraulic actuation-based robot provides the second-slowest laser steering speed.

### 3.2. Free Beam Laser Steering Robots

Free beam is an alternative approach to laser steering. This approach typically allows a faster laser steering and delivery of collimated, focused laser beam by rapidly adjusting the orientation of mirrors (Figure 7). Traditional free beam surgical laser devices have been constrained by the large scale size, limiting their use in minimally invasive surgeries. Motivated by such a challenge, free beam-based laser steering robots have been researched to further miniaturize the robots while providing more than 1000 mm s\(^{-1}\) laser steering speed, as illustrated in Figure 10. To develop these robots, piezoelectric actuators have been utilized to control the mounted optical mirrors. The characteristics of piezoelectric actuators such as small size, high accuracy, and high speed allowed to miniaturize the laser steering system while providing high positional accuracy and laser steering speed. As a result, the size of recently developed free beam style robots has become comparable to that of the fiber steering robots. As listed in Table 2, free-beam-style robots of 6 mm in size (without a camera system and an illumination channel) have been demonstrated. Although this size is still larger in comparison to the size of robots using the fiber steering approach, which are around 1–5 mm, it allows deploying the robot through natural orifices and steer the laser similarly to traditional larger free beam-based laser devices. In terms of the speed performance, the free-beam-style robots outperform the fiber-steering-style robots, as illustrated in Figure 10.

To achieve such performance, various types of piezoelectric actuators, i.e. rotating, linear, and bending, were used. Each type of actuator produces a different output motion when voltage is applied. To start with, a rotary piezoelectric actuator produces the shaft rotation motion by vibrating the stator using a piezoelectric material. Using the spinning motion, Patel et al.[84] rotated a Risley prism. The Risley prism consisted of two wedge prisms and each prism refracted the laser beam to a specific angle. By independently revolving each wedge, the laser beam was steered around the optical axis. However, at the same time, the use of Risley prisms created blind spots within the workspace where the laser beam could not be directed. In addition, many orientations of prisms steered the laser beam toward the center and periphery of the workspace and not in-between.

Similarly, linear motion-based piezoelectric actuators have also been used to maneuver optical components. Rabenrososa et al.[85] and Renevier et al.[86] used a set of linear piezoelectric actuators to steer a laser beam for vocal fold surgery. Their robot included a stereo vision system, fiber bundles with a gradient of index lenses, and a laser steering mechanism. The laser steering mechanism consisted of a pan-tilt mirror actuated by linear piezoelectric motors to change the direction of the laser (Figure 9h). Using the mechanism, a two degrees of freedom laser steering robot was developed with a small package size (18 mm) and a large range of motion (>20 mm × 20 mm). The range of motion was large enough to cover the average size of a man’s or woman’s vocal fold. Although this mechanism successfully miniaturized the free beam style laser steering mechanism, its design required the use of a fixed external prism to guide the laser toward the mirror. Moreover, the overall size of the system depended on the size of the commercially available piezoelectric actuator, making further miniaturization challenging.

On the other hand, the bending type of piezoelectric actuators allowed the development of a different type of laser steering mechanism. Bothner et al.[87] applied microfabrication techniques and bending piezoelectric actuators to manufacture a slider-crank-based laser steering robot. For this robot, two slider-crank mechanisms were used to steer the laser beam. The mirrors were attached to the crank side of the transmission mechanisms, and a quasi-linear motion was generated using the piezoelectric actuators to actuate the sliders. As shown in Table 2, the laser steering speed surpassed the speed of commercially available free beam-based laser steering devices, approximately 200 mm s\(^{-1}\).[90] Improving upon their previous work,[87] York et al.[71] enhanced the performance of their robot by adopting a different mirror design. As shown in Figure 9i, they adopted a three-mirror
design, one fixed and two movable mirrors and enhanced several aspects of the robot, as described below. First, the overall size of the robot (6 mm) is smaller than that of the robot with the two mirror design (11 mm). Second, the three-mirror design maintains the focal distance of the laser despite changing the orientation of the mirrors, which is not possible with the two mirror design. Lastly, they increased the range of motion to $18 \times 18$ mm and the laser steering speed to $3900 \text{ mm s}^{-1}$, achieving one of the fastest laser steering speeds in the literature.

Overall, recent research in free beam-based laser steering robots has enhanced the performance of robots using different types of piezoelectric actuators and transmission mechanisms. However, there are remaining technical challenges to further advance toward a clinically applicable laser device. First, further investigation will be needed in implementing and testing appropriate optical components for the delivery of high-power laser beams listed in Table 1. For each type of procedure, laser sources with different wavelengths and power would be needed. As a result, appropriate optical components would be needed for each laser source. Such requirements would need to be satisfied to utilize the robot as a surgical laser steering robot. Second, electrical safety related to the piezoelectric actuators, a common type of actuators for free beam-based laser steering robots, will need to be further investigated. The piezoelectric actuators could potentially raise safety concerns because they generally require high actuation voltage ($0–200$ V). Lastly, the durability and sterilizability of these robots will need to be further improved by fully enclosing the robot to protect the embedded actuators and transmission mechanisms.

### 3.3. Opportunities and Outlook

For the past decades, many researchers successfully developed laser steering robots with various types of actuators. A common goal of these efforts has been to increase the speed of the robots in steering the laser beam while miniaturizing the overall size to millimeter scale to deploy robots in a minimally invasive manner. In the context of laser surgery robots, the speed and size of these systems are two important parameters that directly impact their clinical outcomes and usability. As illustrated in Figure 10, the use of piezoelectric actuators and magnetic actuators further enhanced performance of both fiber steering and free beam-based laser steering robots. With those actuators, it has been possible to increase the speed up to $3900 \text{ mm s}^{-1}$ and maintain the overall size in the millimeter scale.

Despite the aforementioned enhancements, there are remaining gaps that need to be addressed to further expand the clinical adoption of robotics in laser surgery. To start with, the clinical feasibility of the proposed robots could be further enhanced by exploring parameters that affect the laser–tissue interactions. In many of the presented works, the robots included surgical lasers and focused on controlling the position of those laser beams using mechanical systems. However, the laser–tissue interaction and clinical outcomes not only depend on the speed of the laser beams but also on other parameters like interacting biological tissues, laser wavelength, laser power, and level of focus. For example, controlling the focus of the laser beam, i.e., changing the beam diameter, is an important aspect of laser surgery because it essentially controls the strength and spread of thermal damage to the surrounding tissues. As shown in Table 1, many of the presented works currently do not have such a feature. Thus, the incorporation of features like automatic laser focusing could further improve the clinical outcomes with the usage of the robots because this would indeed avoid manual control of focus and give more options in terms of tuning laser–tissue interaction (i.e., incision, ablation, and coagulation), depending on the level of de-focus.

In the long term, advancements in soft materials would further expand the capabilities of robots for laser surgery. Various characteristics of soft materials (i.e., flexibility, stretchability, biodegradability, biocompatibility, and optical parameters) can be further tuned to develop more optimized laser surgery robots. As recently demonstrated, several groups have shown that actutable soft materials (i.e., electroactive polymers and ferromagnetic polymers) can serve as essential components in developing soft, submillimeter scale laser steering robots. To confirm their practicality, future studies should focus on investigating the biocompatibility and feasibility of these materials as this needs further investigation. For example, magnetic actuation has been showing promises in terms of miniaturizing robots by embedding magnetic microparticles within the robot’s soft body and controlling it with off-board magnets. However, the implementation of neodymium magnet (NdFeB) needs to be carefully considered because NdFeB is known to be cytotoxic due to its corrosive characteristics. Thus, current magnetic actuation methods may fail to meet the biocompatibility requirements of minimally invasive robots, and this may ultimately affect their clinical use and marketability. Likewise, the feasibility aspects of the robots would need to be evaluated by reviewing the surgical requirements, such as the ability to use the robot with other surgical instruments and the capability to steer within a given amount of space.

### 4. Control of Laser–Tissue Interactions

A natural extension of current research in robotic laser surgery is to explore the automation of laser actions. Laser cutting can be difficult to control for many physicians because of a lack of adequate feedback options available to them. Lasers operate in a contact-less fashion, making it impossible to use one’s sense of touch to feel the cutting depth. Furthermore, as we have seen earlier in this paper, lasers can cause thermal injuries whose occurrence can be difficult to anticipate or control given the lack of temperature feedback.

#### 4.1. Challenges of Manual Laser Control

To better illustrate the challenges faced by physicians, it is useful to refer to an example of an actual operating setup. The photographs in Figure 11 illustrate the standard surgical set-up for transoral laser microsurgery, i.e., the surgical laser treatment of the tumors of the voice box. These procedures use a laryngoscope to enable visualization of the surgical site, i.e., a metal tube-shaped device that provides direct line of sight into the larynx. Because of the minuscule size of the anatomy contained in the voice box, physicians perform surgery with a
microscope. Laser incisions are performed manually moving the laser beam by means of a mechanical joystick and activating it through a foot pedal. As we have seen earlier in this paper, state-of-the-art laser aiming systems may offer the automatic execution of preprogrammed laser scan patterns through piezo-actuated fast-steering mirrors. Yet, the scanning mirrors commercially available still retain a manual element, i.e., they rely on the use of the traditional joystick to position the scan pattern on the desired incision line.

From Figure 11, note how far away the physician is from the surgical site. The standard operating distance for these procedures is 400 mm. With the operating physician being so distant from the site of the operation, vision is the only feedback modality that can be used to monitor and control the effects of the laser on the tissue. The incisions performed in these procedures can be so superficial (typically smaller than 1–2 mm) that perceiving their actual depth is a challenge, even under microscope magnification, as it was found in Ref. [89]. Clinically, incision accuracy is obviously of fundamental importance: penetrating too deep into the tissue may damage the vocal muscles (i.e., the muscles that control the motion of the vocal folds) and impair a patient’s ability to speak again.

A second, not less important challenge, is represented by the lack of thermal sensing. As we have seen earlier in this paper, laser cutting is fundamentally a thermal process, and excessive heating can create burns and other hypertrophic scars. Human vision alone does not provide an adequate way to prevent the onset of thermal injuries, as the visual cues that signal thermal damage—typically changes in the tissue color—only appear only after the damage has already occurred.

4.2. Toward the Automatic Control of Laser–Tissue Interactions

In order to assist physicians and improve current outcomes, researchers have focused on providing both enhanced perception and control to physicians. One area that has received attention is providing guidance to physicians to maintain the laser focus during a cutting procedure. If the laser focus changes during a procedure, a physician may be distributing heat over a larger area causing unseen thermal damage. Kundrat et al. have designed a system to provide feedback to a user manually focusing the laser.[64] During a laser procedure, the tissue depth will change causing the laser to become out of focus. Dynamic focusing systems have also been developed that use stereoscopic vision[90] and varifocal mirrors.[91,92]

Multiple research teams have also explored methods to monitor and control laser ablation,[10,59,93–96] with the overarching objective of enhancing the accuracy and safety of laser surgeries. A common approach is to implement laser ablation as a CNC-like cutting process, as illustrated in Figure 12: repeated laser pulses are used to progressively remove material from the tissue surface until a prescribed ablation volume is created.

For this approach to work, the laser ablation process first needs to be modeled; i.e., it is necessary to model the volume of tissue removed by each laser pulse, as well as the thermal effects created by the laser, so that the laser delivery can be properly planned and controlled. Building such models is not straightforward: the way in which laser light interacts with human tissue strongly depends on the tissue-specific composition, including, among other things, the amount of water and blood content.[8,18] These factors are well known to vary significantly among different types of tissue and even among...
specimens of the same tissue type in different individuals.\[12\]

Prior research has successfully produced models that work for specific types of tissue (e.g., fresh ex vivo cortical bone\[94\] or muscle tissue\[10\]), which we review in the following subsections.

4.2.1. Temperature Control

Control of the tissue temperature is key during a laser procedure. As we have seen earlier in Section 2, while a temperature of 100 °C is required to vaporize tissue, temperatures well in excess of this value can result in carbonization. Temperatures below 100 °C can also create irreversible thermal injury in the form of coagulation, which occurs when the tissue temperature is elevated above 60 °C, is marked by visible discoloration (blanching) of the tissue, and it is associated with cellular death.

In principle, one would use the thermal model described by Equation (8) to synthesize a controller capable of regulating the tissue temperature during laser exposure. Unfortunately, this

\[ h(t) = \frac{1}{\rho c} \int_{t_0}^{t} \frac{Q(t')}{A} dt' \]

where $h(t)$ is the tissue temperature, $\rho$ is the tissue density, $c$ is the specific heat capacity, $Q(t)$ is the laser power, and $A$ is the laser area.

Figure 12. CNC-like laser cutting of tissue. Repeated laser pulses are planned to remove a prescribed volume of tissue. Laser application is simulated in software prior to actual execution of the plan. This approach was demonstrated in bone tissue (a,b) and more recently in soft tissue (c,d,e). a) Reproduced with permission. Copyright 2015, IEEE. b) Reproduced with permission. Copyright 2008, IEEE. c) Reproduced with permission. Copyright 2018, Wiley Periodicals. d) Reproduced with permission. Copyright 2010, Springer Nature. e) Reproduced with permission. Copyright 2017, World Scientific Publishing Company.
approach is made challenging by the fact that solutions to this equation can be difficult to solve in closed analytical form.\cite{18} Perhaps even more problematic is the dependence of Equation (8) on the physical properties of the tissue, i.e., the absorption coefficient $\mu_a$, the heat capacity $c_v$, and the thermal conductivity $k$, as these parameters can vary significantly from tissue to tissue and even among different individuals.\cite{21}

Aiming to overcome these limitations, recent work has explored alternative, data-driven methods to model the tissue temperature evolution during laser exposure.

The work in Ref. \cite{97} shows the viability of learning the temperature dynamics of tissue irradiated by a stationary (i.e., not moving) laser beam using nonlinear regression techniques.\cite{97} Using the starting tissue temperature for initialization, the predicted temperature becomes the new current temperature for future time steps. Therefore, using information currently available to a surgeon, i.e., the starting tissue temperature, the laser power, and the laser pulse rate, the tissue temperature can be tracked over time. The performance of the model was contingent on staying within the environmental conditions used for training (constant power and pulse duration below 0.5 s) and was only verified in simulation. This work was later extended in Ref. \cite{98} by modeling the temperature dynamics of the tissue as a Gaussian function. Pardo et al.\cite{99} were able to learn the meta parameter dynamics to track the temporal evolution of the Gaussian temperature profile on phantom tissue. Importantly, these models rely only on the information readily available to a surgeon and do not require any additional equipment or sensors.

Laser procedures generally contain laser scanning motions in order to minimize thermal damage to surrounding tissue. Pardo et al. modeled the surface temperature of the tissue during a laser scanning procedure on ex vivo chicken muscles.\cite{99} They noticed that the change in tissue temperature between successive time steps could be represented as a sum of Gaussians and hypothesized that the temperature profile along the incision could then also be modeled as a sum of Gaussians. This work was verified on ex vivo muscle tissue and gives further evidence that it is possible to learn the tissue temperature dynamics given the same information available to an experienced surgeon.

4.2.2. Tissue Ablation Control

Controlling the laser ablation pattern during a laser procedure is important for a successful outcome. During a procedure, surgeons try to guarantee the complete eradication of diseased tissue while simultaneously minimizing the removal of healthy tissue.\cite{100} While mechanical cutting tools (like a scalpel) provide a physician with tactile feedback during the cutting process, there is no feedback mechanism for laser surgery. Physicians, through extensive training, can learn how to manipulate laser parameters (such as power, energy delivery mode, pulse duration, and scanning frequency) and the laser exposure time to provide adequate cutting.\cite{101,102} There does not exist a standard set of laser parameters and each physician will use different settings depending on their skills and experience.\cite{102} In order to improve surgical outcomes and provide a method to automate laser incision depth, recent work has explored the use of the input parameters defined by surgeons as a means to predict the laser incision depth.

Laser Cutting of Bone: Stopp et al. in Ref. \cite{93} combined a mathematical framework for the ablation depth of a single pulse with preoperative planning and navigation to create precise and safe cortical bone removal. For a given preoperative plan, the navigation control over the laser beam ensures the beam is always focused on the surface of the bone and stays within the desired ablation location. The feed-forward methodology uses simplified mathematical models adapted with an experimentally determined transformation to model the ablation depth of a single laser pulse to guarantee the desired region is ablated. With this feed-forward input, Stopp et al. were able to create the desired cavity depths with less than 1 mm of error.

Burgner et al. in Ref. \cite{94} also investigate how the number of pulses influences the ablation depth. They assume that the ablation created by a laser pulse will be Gaussian shaped and that the total ablation can be described by the series of pulses. This allows ablation depth to be described as a function of the number of pulses the tissue has received. Their work models the ablation depth as a function of the number of pulses and reports that it is an approximately logarithmic function.

Kahrs et al. planned the volume of ablation based on geometric definitions for single laser pulses. In Ref. \cite{95}, the authors measured the shape of a single ablation using a confocal microscope. By separating the planned volume into 2D layers, they can define the required single laser pulses to produce the volume using overlapping circles. This method requires the understanding of an individual laser pulse, which is highly dependent on the laser parameters, and the tissue being ablated.

Leung et al. applied inline coherent imaging (ICI) in order to measure the ablation depth of hard tissue in real time.\cite{103} This imaging technique allows a user to recognize when the ablation is approaching a subsurface structure and cease laser operation. Using this technique, the authors were able to spare a 50 μm layer of bone before a chosen target in cancellous bone. They reported that while it is common to assume material removal is linear with additional laser pulses, they found that in hard tissue this assumption is not sufficient. The number of pulses to initiate the ablation can vary significantly even on the same bone target and after initial onset, the ablation may stall after penetrating into an inclusion within the bone.

Laser Cutting of Soft Tissue: Fichera et al. in Ref. \cite{10} designed and implemented a feed-forward controller for soft tissue ablation. They found that the relationship between laser exposure time and incision depth can be approximated with a linear function. This function is highly tissue dependent and their work was performed on chicken tissue. By inverting the relationship between exposure time and ablation depth, they created the feed-forward model that would determine the exposure time necessary to achieve a desired ablation depth. By modeling the crater created by a single laser pulse as a Gaussian function (similar to Refs. \cite{94,95}), a volume resection could be performed by superimposing multiple Gaussians and determining the appropriate number of laser pulses. In Ref. \cite{89}, Fichera et al. expanded on the work in Ref. \cite{10} to show that the linear model works for both continuous and pulsed laser systems. The model was also extended to work for scanning motions and the effects that scanning frequency would have on incision depth. While the
previous work is focused on single pulses to create a volume resection, this work used scanning motions that are more common during procedures. A user interface was also created to alert the user to the estimated ablation depth, and user trials showed success in reducing user ablation depth error for scanning motions. Both of these approaches, while showing promising results, require the desired tissue to be modeled before experimentation. This reduces their ability to generalize to different tissue and perform well on tissue that is highly heterogeneous.

Acemoglu et al. investigated how energy density and the number of passes along a cutting path affected the ablation depth. While previous work demonstrated a relationship between exposure time and ablation depth, Acemoglu et al. found that both the energy density also had a linear relationship with ablation depth for a fixed power. Changing the laser power but keeping a constant energy density would result in a different ablation depth. They also showed that for a low enough power, there is a linear relationship between incision depth and the number of passes of the laser fiber. This work improves upon the work done by Fichera et al. by allowing arbitrary laser motion and not restricting laser motion to a single direction.

Ross et al. performed the first experiments with a prototype designed for automated tumor resection via laser ablation on a porcine specimen. They used feed-forward open loop control to perform the tissue resection. While their system could perform the resection faster than a human surgeon, they were less accurate. The inaccuracies were caused by heterogeneity in the tissue that the system was unprepared for. The authors did have a real-time surface profiler to measure the depth of the ablation but were not able to use that information for a closed loop controller. Like the previous work, this work is also a feed-forward methodology. However, the authors demonstrated the ability to measure the tissue resected during the procedure. This would allow for future work to create a closed loop controller to control tissue ablation depth.

4.3. Open Challenges

Prior work has successfully developed data-driven laser ablation models that work by modeling the effects a laser pulse has on a known tissue target. This is an important assumption because the optical properties of the tissue will influence the laser ablation dynamics. In a realistic surgical setting, the optical properties of the tissue may not be known a priori, even if the type of tissue is known. The optical properties can vary significantly from tissue to tissue and even the same type of tissue may contain large variations depending on the circumstance. Furthermore, it has been shown that during laser exposure, the optical properties of the tissue may change in response to the laser. To make automated laser ablation viable in a realistic setting, we believe that it will ultimately be necessary to make robots able to identify the optical properties of the tissue being operated on, so that an appropriate ablation plan could be formulated. There is an interesting analogy between this problem and the identification of the mechanical properties of human tissue, which has been extensively explored in previous studies. In these studies, by mechanically perturbing the human tissue and observing its response, these groups were able to identify the tissue's mechanical properties. In order to identify the tissue's optical properties, the material would instead be excited by a laser pulse and its thermal response would be measured. Arnold and Ficher have demonstrated, in simulation, that it is possible to identify the absorption and scattering coefficients of tissue by measuring the temperature change created by a known set of laser inputs. The authors simulate the thermal evolution of tissue using Equation (8) while it is being exposed to repeated laser pulses. Using a variant of the ensemble Kalman filter, they are able to track time-varying optical properties and temperature beneath the tissue surface based on the surface temperature. By monitoring the optical properties of the tissue during a laser procedure, the laser inputs can be adjusted to maximize cutting efficiency and minimize damage to healthy tissue.

5. Conclusion

In modern surgery, lasers have become a widespread tool energy delivery device to perform various types of surgeries. Their abilities to cut tissues with micro-scale precision, minimize surgical trauma, provide hemostasis, and avoid suturing result in unique advantages over traditional surgical tools and end effectors, such as decreased scarring, reduced operation time, faster healing, less inflammatory response, and less intra- and postoperative pain. However, current laser devices alone leave several burdens to surgeons. Conventional laser devices require complex and dexterous manipulation and do not provide any penetration depth information, requiring extensive training to adequately perform laser-assisted surgeries.

To address the limitations of current surgical lasers, researchers in various fields developed robotic platforms that include different actuation strategies for laser steering and computational models that estimate laser ablation volume based on the laser–tissue interaction. The developments from these research groups have been providing outstanding preliminary results toward robotic platforms for laser surgery. However, there are several short-term and long-term challenges that need to be addressed to progress toward the next-generation robotic systems for laser-assisted surgery. In the short term, problems associated with transferring the developed robotic platforms to real clinical environments should be addressed and effectively transition from laser procedures performed manually to robotic-assisted laser surgery. These will involve different design parameters including: selection of appropriate actuation methods, electrical safety associated with actuation methodologies, appropriate material selection to ensure biocompatibility and durability of mechanical structures, and development of software and hardware interfaces. In the long term, the advancement of miniaturized actuators, investigation of materials for mechanical structures, and development of laser–tissue interaction models for various types of tissues would be needed to further expand the applications of robotic laser surgeries. These advancements ultimately would automate various types of laser-assisted surgeries, including soft and hard tissue ablation and micro- to nanoscale surgery.
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Conflict of Interest
The authors declare no conflict of interest.

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