ABSTRACT

MOHAMMADBAGHERPOOR, HAMED. Designing and Testing a Closed-Loop Magnetically Actuated Laser Scanning System for Tissue Ablation. (Under the direction of Edward Grant.)

The research here dealt with improving the control of biological tissue ablation over currently available systems being used for this task. Two prototype laser scanner systems were designed, fabricated and tested in biological tissue ablation experiments. In prototype 1 was an endoscopic tool design that controlled the displacements of a flexible fiber optic cable through controlling the current to four miniature electromagnetic solenoid coils individually, in a N-S-E-W configuration. The torque necessary to displace the flexible fiber optic cable was generated by the interaction between electromagnetic fields. That is, by placing small permanent magnets around its circumference of the fiber optic cable and by varying the current in the solenoid coils a torque was generated. Prototype 1 had fixed optic lenses and included a novel photo-detector sensor to give the endoscopic tool feedback control. Experiments, on wood and biological tissue with prototype 1 gave mixed results. Complex geometric patterns were reasonably accurately ablated in these experiments but the pulsed laser used in the scanner system damaged tissue at tissue sites surrounding the line of the geometric pattern. Based on the test results from prototype 1, a prototype 2 laser scanning system was produced. The prototype 2 endoscopic tool was designed to be a modular construction, scalable and with adjustable optics and feedback control. A completely new prototype 2 laser scanning system was custom designed and assembled around a Lumics Diode continuous wave laser. This prototype 2 system successfully generated the geometric ablation patterns in the biological tissue ablation experiments. The biological tissue results obtained with prototype 2 did show an improvement over prototype 1. However, in order to refine and optimize the design of prototype 2 it was modeled in COMSOL. The conclusion drawn from the physical experiments undertaken and the simulated results obtained from the COMSOL model of prototype 2 are that laser scanning system designs require improved optic lenses, ultra-fast laser pulsed laser systems.

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Designing and Testing a Closed-Loop Magnetically Actuated Laser Scanning System for Tissue Ablation

by
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DEDICATION

To my wife, Mehrnaz, who constantly support and encourage me during Ph.D. challenges and life.

To my son, Daniel Pasha, which made me a stronger, better person in my life and happier than I could ever imagine. I love you to the moon and back.

BIOGRAPHY

Hamed Mohammadbagherpoor was born in 1987 in Neyriz, Fars, Iran to parents Masoumekhatun and Mohammadreza. He attended the Shiraz University of Technology in Iran in 2005 and received his Bachelor of Science in Control Engineering. In 2012 he continued his education at the University of Tehran and received a Master of Science in Control Engineering. In the Spring of 2016, he moved to Raleigh, NC, USA to pursue his Ph.D. degree in Electrical Engineering in Robotics and Mechatronics at North Carolina State University. He has been fortunate to join the CRIM lab and started his education under the supervision of Professor Edward Grant.

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CHAPTER

1

INTRODUCTION

1.1 Background

The first visible light laser system was invented by Theodore Maiman in 1960 [Mai60b]. This first laser system used a ruby crystal. Over the decades since, many technologies have been used to improve the quality and power of laser light so that they could be used wide range of applications. This extensive research has led to the invention of CO₂ lasers, Nd:YAG(Neodymium:Yttrium Aluminum Garnet) lasers and Holmium lasers. In particular, lasers that have been widely used in medical applications [Pol70], particularly in medical ablation surgeries. It is the control of the laser light beam in tissue ablation surgeries that is the focus of this dissertation and in particular the introduction of feedback, so that such surgeries can progress towards partial automation as opposed to the teleoperated laser systems that is used by surgeons today. The most common application of CO₂ lasers in surgery is the cutting and ablation of biological tissue. Their use, to a large extent, is based on the degree of coagulation experienced during a surgical procedure. Today, lasers are used in many medical applications, including: (1) dermatology, (2) ophthalmology, (3) gynecology, and (4) laryngology.

Figure 1.1 Larynx laser ablation. The optical fiber and the camera lens are placed at the left and right side respectively [Hu17]-[Thr].

In laryngeal surgeries, lasers are used for: (1) laryngeal lesion operations, (2) airway stenosis, and (3) laryngeal carcinoma. In a typical laryngeal ablation surgery, such as the removal of a tumor from a larynx, a free-beam CO₂ laser is used in tandem with the surgical microscope to facilitate the tumor removal [Str72; Str75]. Fig.1.1 shows a general view of a typical laryngeal surgery. As can be seen, a laryngoscope is placed into the larynx directly below the surgical microscope. The surgeon then controls the positioning of the CO₂ laser beam on the tissue by manipulating by teleoperation, i.e., by using a small joystick. This approach to laryngeal surgery is highly effective, precise, with good outcomes and so it is popular with surgeons worldwide. However, the control of the micromanipulator, via teleoperation, remains a kinematic challenge for the surgeon [Sim18; Avg16].

The manipulation of the micromanipulator becomes uncomfortable over time, and in particular, when the area of tissue to be ablated is small, and precise tissue ablation is critical to the quality of the outcome. In recent years, researchers have worked continuously on the development of new laryngeal surgery technologies that can: (1) improve the quality of tissue ablation, and (2) reduce the size of laser scanning systems, e.g., the endoscopic tool [Mat19]. The introduction of robotic systems into laser assisted surgery, as well as the integration of a host of other technologies from physics and engineering, e.g., cameras, ultrasound sensors, optical fibers, and lasers, has benefitted medical procedure outcomes [Sim18]. These days, a common application of these integrated technologies in medical surgery is in laparoscopic surgery and microsurgery, mainly through the design and development of new endoscopic tools. But, endoscopic tool design challenges remain, particularly

so in the precise control of their actuator systems [CV20]. Accuracy is needed in order to conduct laser microsurgery based on laser raster scanning, i.e., a move towards partially automating tissue ablation surgery, which is a considerable surgical challenge because of the: (1) small areas that are commonly ablated, and (2) confined workspaces within the human body cavity, in which a raster scanning system is required to function. Typically, a laser actuator control system, i.e., the system that navigates and localizes the laser light beam, is located outside of the patient, but it still requires a direct line-of-sight into the surgical workspace [Ace17]. This is a major reason why lasers systems are being used in surgical procedures that are being conducted on delicate organs inside the human body cavity, e.g., eyes or organs that require up close and accurate and precise laser beam navigation and localization. For example, eye and throat surgeries are typical examples of such procedures [Gia12]. Properties inherent in laser systems make them suited to surgical procedures including both monochromaticity and collimation. These two laser systems properties produce high-intensity energy laser beams, i.e., high power in a narrow beam, which can in turn ablates tissue and organs precisely [Sol61]. The first application of a laser system in medicine was carried out in 1963 when a laser was used for a retinal tear surgery [Cam09]. In the early 1980s, a laser system was applied in a wide range of orthopedic procedures [Ger81]. In these early operations, both continuous lasers and long-pulse lasers were used in ablation procedures, but they were shown to cause both carbonization and tissue fracturing around the tissue margins. The researchers in [Kro89; Nie07] showed that the level of laser irradiance had a major impact on the outcomes of laser ablation procedures, i.e., an acceptable outcome of laser ablation procedures was achieved when a low average energy laser light was focused on a small area. The laser system introduced by Myers [Ver12] had the ability to cut bone precisely with little carbonization and mechanical stress. [Hen12] introduced CO₂ lasers into medical procedures but their initial low light energy levels gave inconsistent results. When the light energy levels of CO₂ lasers was increased, i.e., magnified, by introducing carbon dioxide molecules into the device, tissue ablation experiments with CO₂ scanning lasers showed improvements in the quality of laser microsurgery [Hen12]. Additionally, scanning laser systems allowed for laser ablation procedures to be carried out in a fast and efficient way, with a significant reduction in levels of thermal damage to any adjacent healthy tissue [Hen12]. The benefits obtained through using lasers in soft tissue surgeries have included: (1) improved post-operative functionality, (2) decreased morbidity (disease), (3) better homeostasis (coagulation, changing blood from a liquid to a gel), and

(4) minimal peripheral tissue injury [Sol07; Des08]. Integrating laser endoscopic scanning tools and robotics has helped improve the accuracy of the surgeons' performance, particularly in the laser ablation of tumors [Che18; Zha20]. Endoscopic tools are readily available medical devices for use with laser systems. In combination with optic fiber cables, these endoscopic tool systems are small and accurate in practice, i.e., they are well suited to microsurgery procedures. The actuation control of the laser in these endoscopic tools is through a human-computer interface, i.e., the surgeon's control interface, a joy-stick which was mentioned earlier. This interactive user interface is located outside of the human body, because of its size. This size limitation, led to early research into the design, manufacture, and control, of a microsurgery laser ablation system. This early prototype system proved that an ablation procedure could function inside the human body, see Giallo and Grant [Gia08]. It was proved by [Gia08] that the precise control of such a device increased the quality of microsurgery. However, the control of microsurgery laser devices also brought new technical challenges. This research was further extended by [Ace17] in the European Union funded μ RALP project.

The research here is focused on extending the μ RALP research even more, to give fine control of the automated fiber-coupled magnetic laser scanning system that was built as a test-bed. The research here moves the automation of endoscopic laser scanning microsurgery closer. In this dissertation, a new system with four miniature solenoid coils was designed to control a fiber optic cable, similar to the μ RALP system. However, in this research the individual solenoid coils that surround the optical fiber cable, in a N-S-E-W configuration, are controlled individually. In the μ RALP project these coils were controlled in a NS-EW manner. These individually actuated N-S-E-W coils operate in tandem with a set of permanent magnets that are attached to the free end of the optical fiber cable. Together they form the coil actuator control system. An electronic driver interface board was designed and fabricated to accurately control the current being supplied to the coils. By supplying current to the coils an electromagnetic field was generated around the optical fiber. The interaction between the permanent magnets and the solenoid coils produces the torques and forces required to control the navigation and localization of the optical fiber cable. The major goal of this research was to ensure that the assembled test-bed was a closed-loop computer-driven system based on the miniature solenoid coils and electromagnetic fields. To achieve this goal, and improve overall laser light beam control, the output pose of the laser beam light was measured by a

photo-detector sensor and fed-back to the controller, i.e., a computer, thereby achieving closed-loop control. By closing the loop, an appropriate control signal was generated. This control signal was further added to a feed-forward control signal, and this improved the tracking accuracy of the laser light beam. In experiments, this test-bed accurately controlled the optical fiber cable and proved that the laser beam could accurately and precisely generate any desired geometric pattern desired by the user. The advantages derived from using this actuation approach include (1) non-contact actuation, (2) high-speed scanning, and (3) high resolution.

1.2 Thesis Structure

This thesis is divided into seven chapters: (1) Introduction, (2) Interaction between laser beam and tissue, (3) Electromagnetic sensors and actuators in biomedical engineering, (4) Magnetic Laser scanner system design (Prototype 1), (5) Experimental results (Prototype 1), (6) Laser scanner design and evaluation (Prototype 2), and (7) Conclusion. Chapter 2 includes a detailed overview of laser and tissue interaction, background and the application of laser system in biomedical. Chapter 3 describes the applications of the electromagnetic sensors and actuators in biomedical engineering. An electromagnetic based proximity sensor was designed and developed to measure the loosening of total joint replacement. Chapter 4 covers the design and fabrication of the first electro-mechanical prototype1 of the magnetically actuated laser scanner system. Chapter 5 presents the primary experimental results of the first designed prototype 2 system. Different scenarios were considered to evaluate the assembled system. In chapter 6 a scaled-down magnetically actuated laser scanner system with the experimental results are designed and discussed. Finally, chapter 7 presents conclusions on the research conducted.

CHAPTER

2

LASER TISSUE INTERACTIONS

2.1 Introduction

Industrial applications of laser scanning systems have included cutting, welding, and 3D printing [Hat12]. And, laser-scanning systems are also used extensively in medical applications, for: (1) medical imaging, (2) welding together blood vessels, (3) diagnoses, and (4) laser microsurgery [Kat93]. A major use of laser systems in medicine is medical microsurgery and soft tissue ablation. Today, in microsurgery, endoscopic tools and laser systems are widely used, initially their use was an attempt to improve the quality of ablation surgeries [Boy18; Har17; Žem18]. However, the design of the actuators used to control laser-based endoscopic tools remains a challenge, because of the: (1) complexity of the human body's organ systems, and (2) scale of the actuators themselves. As mentioned in Chapter 1, section 1.1, large scale of today's endoscopic tool control systems has them placed outside the human body [Žem18]. These systems, in turn, control the actuators embedded inside the endoscopic tool. By deflecting the laser beam emitting from the fiber-optic cable, the laser scanning system ablates tissue. This chapter provides an overview of the: (1) medical applications

for tissue ablation, (2) effects of laser light on tissue, and (3) laser scanning systems.

2.2 Medical Laser Scanning Systems

Currently, laser scanner systems are used in a wide range of medical applications, e.g., coagulation, welding blood vessels, microsurgeries, and diagnosis [Gho11]. Laser scanning systems are being used as an alternative to medical imaging systems such as computed tomography (CT) and magnetic resonance imaging (MRI). However, it is difficult to recognize many internal parts of the human body when using a 3D laser scanner. 3D laser scanning systems are also being used for scanning and constructing a 3D model of the outside of the human body. Medical practitioners use the outside scanning approach to examine patients with burned skin, because the surgeon requires a 3D map of the patient's body in order to plan the accurate removal of the damaged tissue. In such cases a laser Doppler scanning system has proved accurate for measuring the depth of burned tissue sites [Ven14]. In 1980, [She80] used a laser scanning system to improve the resolution of a scanning system, by introducing vibration and optical microscopy into the system. This enhanced the contrast of the images gathered by a laser Doppler scanning system significantly. Thus, enhancing the optical properties of tissue in this way improved patient diagnosis [Kat93]. More recently, MEMS devices have been designed and used for in vivo imaging. [Yel06] shows how an endoscopic laser scanning system can be used for 3D imaging. In this approach, the actuator of this laser scanner was controlled by a d.c. motor and a galvanometer.

2.2.1 The Range of Laser Systems in Medicine

Laser systems are categorized by their: wavelength, temporal operation, quantum transition, pumping, and active material. Laser systems, including CO₂, Erbium-doped: yttrium aluminum garnet (Er:YAG), Neodymium-doped, yttrium aluminum garnet (Nd:YAG), and Holmium: yttrium-aluminum-garnet (Ho:YAG) have all been used for endoscopic surgeries.

1. CO₂ Lasers: in this laser system, light is magnified by introducing carbon dioxide molecules; which produces an emitted laser wavelength of 10.6 μ m. With this wavelength, a CO₂ laser can cut materials, weld metals, and carry out ophthalmology surgeries, because CO₂ irradiation is

absorbed by biological tissue. However, there is commonly vaporization and heat damage to the surrounding tissue [Rob70]

2. Er:YAG Laser: has a laser light wavelength of 2940 nm . The irradiation, in this case, is 10 times larger than that of a CO_2 laser, leading to a significant amount of water absorption. Due to the smaller wavelength and low thermal damage, this laser system is used in medical procedures such as periodontics surgery [Per98; Bad06]
3. Ho:YAG Laser: has a wavelength of $2.12\mu\text{m}$, which again leads to an associated higher water absorption. This laser can vaporize biological tissue, and cut both soft and hard biological tissue. This laser is mostly used for prostatectomy and tumor ablation procedures [Shi98]
4. Nd:YAG Laser: has a wavelength of $1.064\mu\text{m}$. Water absorption from this laser system is less than that of any other laser device. Also, absorption in oxyhemoglobin is 10 times greater than water. This laser can only be used for the ablation of superficial layers, e.g., it is effective for human leg vein ablation [Gaš01].

The invention of the CO_2 laser provided medical practitioners with new, and precise techniques for soft tissue ablation [Pat64], and ablation experiments on vocal cords showed that a CO_2 laser ablated tissue with reduced bleeding [Jak72]. Compared to conventional biological tissue ablation techniques with lasers, adding a microscope to the laser: (1) improved the precision of tissue ablation, (2) minimized patient post-operative pain, (3) decreased the operation cost, and (4) reduced morbidity [Vau78; Whi91]. Early research demonstrated that these laser ablation systems could replace radiation therapy techniques and that they could replace radiation therapy for removing laryngeal cancers [Sko75; Rub10]. In 1991 Milford [Mil91] demonstrated that lasers could remove laryngeal cancers precisely and with a significant reduction in the negative side effects commonly experienced with radiation therapy, e.g., tissue scarring. The use of lasers systems for tumor ablation is often accompanied by thermal damage and tissue carbonization. In [Mil91] the authors proposed controlling laser movement as a means of controlling the ablation process. This led to a reduction in tissue carbonization and reduced thermal damage on healthy tissue surrounding the ablation site. These new laser scanner systems have improved the results of biological tissue ablation and they have minimized the damage experienced by healthy tissue [Rem04; Rem08]. The introduction

Figure 2.1 a) Free-Beam Laser System [lum] b) Fiber Coupled Laser System [lum].

of piezoelectric actuators into endoscopic tool designs led to such a laser scanner successfully ablating vocal cord tissue [Fer14]. A separate research program proposed a laser scanner system with a wavelength of light that was controllable. This controllable wavelength of light system also ablated vocal cord tissue in 2D successfully [Tsi09]. In microsurgery two types of laser scanner systems are used for biological tissue ablation: (1) free-beam systems, and (2) the fiber-coupled systems, see Figure 2.1. The laser system in Figure 2.1 uses a CO₂ laser, because the CO₂ laser can be used in both a free beam system design, or in a fiber-coupled laser system design. The advantages of using laser microsurgery ablation are: (1) high accuracy, (2) time and cost-effectiveness, and (3) eliminating human error. The disadvantages are: (1) burnt tissue, and (2) the need for a highly-skilled, highly trained, medical practitioner to conduct the surgical procedure [lum].

1. Free-Beam Laser: This system comes equipped with scanning tools. These tools help to eliminate tissue carbonization and give accurate tissue ablation. Free-beam laser scanning systems are used for ablation procedures conducted over a long distance, e.g., hundreds of millimeters [Cha17a].
2. Fiber-Coupled Laser: Fiber-coupled laser systems are more flexible laser scanning systems than their free-beam laser counterparts. They are highly effective in endoscopic tool laser surgeries that require the laser energy to be applied to specific organ sites inside the human body cavity, and in close proximity to the tissue ablation site. Due to its size, a fiber-coupled laser system is ideal for minimally invasive surgery [Cha17a]. In this dissertation, a new fiber-

coupled laser scanning system was designed, fabricated and made into a test-bed. This new flexible fiber-optic cable laser scanning system was incorporated into a custom-designed endoscopic tool. The laser scanning system was controlled by: (1) controlling the current in an electromagnetic field of four solenoid coils, (2) introducing a light sensor and beam splitter into the endoscopic tools optics, i.e., to provide feedback to the systems controller (see Chapter 4, section 4.3).

Magnetic catheters have also been used in surgical procedures to assist in procedures being carried out in confined operating spaces inside the human body cavity. A magnetic catheter guidance system outside the body navigates the catheter inside the body. These systems are used for cardiac electrophysiological ablation [Bos16]. Separately, a coupled mechanical and magnetic actuation system has also used with surgical catheters, i.e., the closed-loop control of the magnetic catheter [Ull14]. Researchers also developed a computerized closed-loop position control system to navigate catheters using permanent magnets. This system was used inside both the heart and body cavity. The catheter system was guided by a magnetostatic field, that was once again generated outside the human body. This magnetostatic field controlled both the direction and magnitude of the navigation [Bos16]. Lastly, a contactless, magnetically actuated, optical fiber cable was proposed for controlling a laser beam using the electromagnetic forces [Ace17]. The advantage of this system was: (1) the simplicity of its control system design, and (2) the speed with which the fiber optic cable could be controlled. However, due to the system being an open-loop control system only, the system required calibration and tuning for every new reference signal. And, the fiber optic cable was controlled by two orthogonally actuated electromagnetic coils, in a coupled N-S, E-W; activation arrangement. The forces generated by the electromagnetic coils interacted with five-ring permanent magnets that were placed on the optic fiber cable. This research was at the core of the research contained in this dissertation.

2.2.2 Medical Applications of Laser Systems

Historically, in medicine, the major applications of laser systems started with their use in ophthalmology and dentistry [MS49; Gol64]. Today, medical laser systems are used in the treatment of cancer, e.g., in photodynamic therapy and thermotherapy. In this section, some of the major

application areas of medical laser systems will be discussed, starting with Ophthalmology.

In *Ophthalmology*, laser systems are used to both diagnose diseases and carry out therapeutic treatments. For example, laser systems are used for the early detection and diagnosis of retinal alteration in cases where the conventional light technology fails to show certain conditions. Meyer-Schwickerath [MS49], the father of light coagulation, was the first to use laser light beams and xenon photocoagulators for retinal treatments, immediately following the invention of the ruby laser system. His work showed that the ruby laser system was a suitable tool for retinal treatment and segment detachment, but that it was ill-suited for closing blood vessels and stopping bleeding. Later it was found that the argon-ion laser system was more effective than the ruby laser in closing blood vessels [Mai60a]. This was due to the argon-ion lasers green and blue light wavelength, because its green-blue wavelength was absorbed by the blood vessels, whereas the ruby laser's red wavelength was not. The argon-ion laser system was used in ophthalmology for lens treatment (Cataract surgery and fragmentation), iris treatment, sclera, and cornea treatment [Zar61; Zwe64; Cam63].

In *Dentistry*, the second major area of application of lasers in medicine, laser systems were used to treat both hard and soft dental tissue. First, in 1964 [Gol64], a ruby laser with a $694\mu\text{m}$ wavelength was used for treating a tooth. However, the results were less than satisfactory, because the tooth was cracked and the laser system caused severe thermal damage. Further attempts on treatment teeth were made using lasers systems with variety of wavelengths. But, thermal damage still resulted and laser systems were deemed unsuitable for tooth preparation. In 1989 an Er:YAG laser, with a $2.94\mu\text{m}$ wavelength, was introduced into dentistry [Hib89]. This laser system had a resonant frequency close to that of water molecules inside of teeth and was applied successfully in dental thermomechanical ablation procedures. Today, Nd:YLF laser systems, with picosecond and femtosecond pulses, are used widely for treating caries and removing caries [Log18]. In addition, laser scanning microscope systems have been applied to precisely determine the volume of the filling material needed for a cavity. Lastly, laser systems in dentistry are used to sterilize an infected root canal and to remove plaque from teeth. So, laser systems are augmenting the performance of conventional dentistry techniques, e.g., the use of anti-microbial chemicals used in sterilization.

Gynecology is a major medical field where laser systems have improved the treatment and removal of new and uncommon cervical tissues. For example, Cervical Intraepithelial Neoplasia

(CIN) must be treated as soon as it is detected, to prevent cervical cancer efficiently. CO₂ laser systems with three different temporal modes (Continuous Wave(CW), chopped pulses, and super pulse) and three focusing modes (partially and tightly focused or defocused) are applied in gynecology treatment. CO₂ lasers, with their low power density, are used for Vaginal effectively for Intraepithelial Neoplasia (VIN) treatment [Sta77; Wan14]. Laser surgery is also an excellent tool to treat a Twin-Twin Transfusion Syndrome for twins inside the womb. In this application, an ND:YAG laser system is the only procedure known that can remove a shunt vessel found between the actual twins. This shunt vessel produces an unbalanced blood supply to the twins.

The first use of a laser system in *Urology* was conducted by Staehler et al. in 1976 [Sta76]. Their procedure used a fiber optic endoscopic tool in concert with an Argon ion laser system. The results were promising and led to a vast amount of research into the application of laser systems in urology [Nie07]. That original medical procedure extended to treat many other urology conditions, including: external genitals, bladder, ureter, and urethra. Three main laser systems have the optimal power and wavelength characteristics to deal with urological conditions, they are: Nd:YAG, Ho:YAG, and CO₂ lasers. So, they have also been used successfully in lithotripsy and prostatic hyperplasia, i.e., to break down bladder stones, remove bladder tumor cells and stenotic material in the lower urinary [Nie07].

Generally, brain and spinal cord surgeries are the most difficult surgeries performed [Ros66]. This is due to their structure, complexity, and accessibility. For example, brain and spine tumors are often surrounded by other vital cells. Research into brain and spine surgery continues, to find valuable, simple techniques, for removing life-threatening tumors with little or no side effects for the patient. Laser systems applied in *Neurosurgery* applications are limited by the type of laser system chosen, and the amount of power used during the surgery [Nie07]. Thermal damage to healthy neuron cells must be avoided at all costs. Both the Nd:YAG and moderate CO₂ lasers are considered the most useful in neurosurgery to cut the tissue, coagulate the blood vessels and vaporize the tumor cells. The less there is of brain-laser interaction the less the amount of post-surgery infection [Nie07].

In 1986 Hussein [Hus86] introduced a new hot tip tool in which a piece of metal is placed at the end of fiber cable. This tool converted all the laser's incident light beam energy into heat energy, which was then used in surgeries for vessel recanalization, *Angioplasty and Cardiology*. Any kind

of laser system can be applied to the hot tip. However, Nd:YAG and CW Argon ion lasers are the most applied lasers in Cardiology. A pulsed XeCl excimer laser is applied in vessel canalization, with resulting low thermal damage to any local healthy tissue [Gru85]. Laser systems are also showing promise in the diagnosis of angioplasty and cardiology. Miniaturized and sensitive laser endoscopic tools are being passed through the blood vessels around the heart and play an important role in diagnosis.

Laser use in the treatment *Gastroenterology* diseases, e.g., ulcers and tumors, continues to show good results. CW Nd:YAG laser systems are applied to coagulate blood to stop bleeding in vessels [Nie07]. Also, laser systems with the high power density and in vaporization mode, are used by surgeons to open the stenoses. If left untreated, stenosis can lead to patients experiencing numbness, loss of balance etc. The operation directs the laser light beam to the identified site through the optic fiber and, with a power density set to coagulate any vessel, removes tumor related tissue. In an attempt to shrink a tumor, the fiber tip is commonly placed inside the tumor, and the heat from the radiated light beam is absorbed by the tumors tissue and converted into heat. The laser's power intensity is increased temporarily during the surgery, in order to coagulate the tumor efficiently [Bad86].

In *Dermatology and Cosmetics* laser surgeries, coagulation and vaporization are the two most common photothermal effects that arise. The laser systems of choice for Dermatology and Cosmetic laser surgeries are Argon ion, CO₂ lasers, and Nd:YAG lasers. As with other medical procedures CO₂ lasers are used for vaporization and the removal of tissue. CO₂ and Argon ion lasers, with their variety of power densities and wavelengths, are used for tattoo removal. The application of Argon ion laser is limited to superficial treatment; such as port wine stain removal, because its radiated beam is absorbed by hemoglobin and melanin [Apf78; Apf79; Nie07].

Lastly, the use of laser systems in *Orthopedics* was the final major medical field reviewed in this literature search. In 1970 a CO₂ laser system was applied to bone healing [Nie07; Fuk10]. The first set of experiments showed that a CO₂ laser had the ability to ablate bone material, but with a high risk of damage to healthy surrounding tissue. In 1993 Forrer et al. [For93] demonstrated that CO₂ laser with a short pulse width, one with 9.6 μ m wavelength and a power density of 15 J/cm², ablated bone in the micron range of thermal effects. This was due to the bone material having a higher absorbance rate at 9.6 μ m compared to 10.6 μ m. And, using a shorter pulse helped to reduce

Figure 2.2 Optical properties of matter [Vog03].

thermal damage to local healthy tissue. Er:YAG lasers with $2.94\mu\text{m}$ wavelength and Ho:YAG lasers with $2.12\mu\text{m}$ wavelength are the other two lasers system types that can be used to ablate both bone and cartilage [Nel89; Gon90; Nus88; Ste90].

2.3 Tissue Interaction with the Incident Light Beam

Understanding the properties of both a laser light beam and biological tissue are the two key factors in conducting successful laser ablation procedures. Knowledge of these two properties helps in: (1) the planning of an ablation surgery, (2) ensuring a quality ablation procedure outcome, (3) providing system efficiency, and (4) minimizing any thermal damage to healthy tissue. Both laser light scattering on the tissue and the absorption of energy by tissue, play a crucial role in all laser ablation procedures. This is because the levels of radiation energy density transferred by the laser to the tissue and the distribution of that energy throughout the tissue during an ablation procedure are not an exact science. Photons can be absorbed or scattered by the tissue in a random manner and in all directions. Two important properties of the tissue must be considered whenever a laser light beam is used in medicine, that is: (1) the absorption coefficient (μ_a), and (2) a low scattering coefficient (μ_s) [Lin97]. A greater understanding of these parameters leads to precise ablations and reduced thermal damage to tissue. Several methods are available to measure the optical properties of tissues, including: (1) integrating-sphere measurement, (2) normal-incidence video reflectometry, and (3) interferometric photothermal spectroscopy [Wil95]. Other properties, e.g., reflection, absorption, and scattering properties of light, are discussed in the following sub-sections.

2.3.1 Reflection and Refraction

Reflection is defined as the amount of surface reflection obtained from an electromagnetic field radiated by a laser system. This surface is the boundary between two different materials. Commonly, the amount of reflected light depends upon the smoothness and the texture of the surfaces involved [Nie07]. Reflection can be categorized as specular reflection and/or diffused reflection. Specular reflection occurs when tissue imperfections are smaller than the wavelength of the incident light, i.e., light reflected equally. Diffuse reflection occurs on rough surface in which light is reflected in all directions. Commonly, diffuse reflection is the most common type of reflection, and understanding its impact on tissue plays a significant role in the laser ablation process [Nie07]. Figure 2.2 shows different behavior of matter in light. Refraction occurs when the speed of the light wave changes at the surface of two media Figure 2.3. Based on Snell's law, the refraction can be formulated as:

$$\frac{\sin(\theta_i)}{\sin(\theta_t)} = \frac{v}{v'} \quad (2.1)$$

where θ_t is the refraction angle, v and v' are the speed of light in different medias. Eq.2.1 can be reformulated based on speed of light in vacuum (c).

$$\begin{aligned} n &= \frac{c}{v}, \\ n' &= \frac{c}{v'}, \end{aligned} \quad (2.2)$$

$$n \sin(\theta_i) = n' \sin(\theta_t)$$

2.3.2 Absorption

When photons penetrate the electromagnetic field surrounding a medium, they are absorbed by that medium and the light intensity of the medium increases. Medium absorption is due to lightwave energy being converted into: (1) heat energy, or (2) energy that causes tissue molecules to vibrate. Light waves with specific wavelengths can pass through a transparent medium, e.g., lenses or a cornea, with low levels of energy being absorbed. However, the ability with which biological tissue absorbs electromagnetic radiation depends on factors, such as the: (1) electronic constitution of the

Figure 2.3 Refraction of light in different tissue medium [Vog03].

atoms making up the tissue, (2) internal temperature of the tissue, (3) tissue concentration, and (4) thickness of the tissue layer. The absorption property of tissue is expressed by Lambert–Beer’s law [Nie07]:

$$I(z) = I_0 e^{-\mu_a z} \quad (2.3)$$

where z is the optical axis, $I(z)$ is the intensity at a distance z from the surface, I_0 is the incident intensity, and μ_a is the absorption coefficient ($c\ m^{-1}$). The inverse of the absorption coefficient is referred to absorption length L :

$$L = \frac{1}{\mu_a} \quad (2.4)$$

The major energy absorption medium in biological tissue is water molecules, followed by macro-molecules, like proteins and pigments. Water molecules mainly absorb in the IR spectrum range, Figure 2.4, while proteins and pigments absorb in the ultraviolet (UV) and visible spectrum ranges. According to Boulnois (1986) [Sin87] the absorption peak for proteins occurs at 280nm . However, the absorption coefficient of water molecules changes suddenly when the wavelength of the light increases from 900nm to 3000nm . Also, melanin and protein have high absorption rates in the visible and UV spectrum ranges. For wavelengths less than 200nm , water light absorption changes rapidly [Vog03]. In some medical laser applications special dyes and inks are used to increase the absorption coefficient of the biological tissue prior to the surgery in order to improve the efficiency of the laser treatment.

Figure 2.4 Optical absorption coefficient of biological tissues [Vog03].

2.3.3 Scattering

Scattering happens when a light beam deviates from a straight trajectory. Commonly this is due to non-uniformities in the medium materials. Elastic and inelastic scattering are the common types of scattering. In elastic scattering, both the incident photons and scattered photons have the same energy levels, i.e., the scattered photons have the same wavelength as the incident light. Rayleigh scattering is considered as a special type of elastic scattering. In inelastic scattering, however energy is transferred between the incident light and the medium material. This results in the scattered photons having a different wavelength to that of the incident light. With inelastic scattering, a laser with a long wavelength can be used to ablate tissue, because some of the energy is absorbed by the biological tissue[Vog03]. Figure 2.5 shows the geometry of Rayleigh scattering. By considering the incident wave, the corresponding electric field can be defined as:

$$E(z) = E_0 e^{(ikz)} \quad (2.5)$$

where E_0 is the original amplitude of the incident beam, z is the optical axis, and k is the propagation vector. The amount of loss will be defined as:

$$I(z) = I_0 e^{(-\mu_s z)} \quad (2.6)$$

Figure 2.5 Scattering of light by tissue medium located at the origin [Vog03].

where μ_s is the scattering factor coefficient. By considering the density of scattering atom for a medium with thickness L we have:

$$\begin{aligned}
 I_s &\sim \mu_s L \\
 I_1 &\sim \frac{\mu_s L}{NL} = \frac{\mu_s}{N} \\
 E_1 &\sim \sqrt{\frac{\mu_s}{n}} E_s \sim NL \sqrt{\frac{\mu_s}{n}} = L \sqrt{\mu_s N}
 \end{aligned} \tag{2.7}$$

where I_s is the intensity scattered by the medium, I_1 is the intensity scattered by one atom, E_1 is the corresponding electric field, and E_s is the total scattered electric field based on μ_s and N . By adding the Eq.2.7 to the Eq.2.5, the complex amplitude at distance z can be calculated as:

$$E(z) = E_0 \left(e^{ikz} + L \sqrt{\mu_s N} \int_0^\infty \frac{e^{ikR}}{R} 2\pi r dr \right), \quad R^2 = r^2 + z^2 \tag{2.8}$$

By changing the variable and neglecting the ∞ boundary due to the finite wavelength, Eq.2.8 will be as:

$$\begin{aligned}
 E(z) &= E_0 \left(e^{ikz} + L \sqrt{\mu_s N} \int_z^\infty \frac{e^{ikR}}{R} dR \right) \\
 E(z) &= E_0 \left(e^{ikz} - L \sqrt{\mu_s N} \frac{2\pi}{ik} e^{ikz} \right)
 \end{aligned} \tag{2.9}$$

by considering $\lambda = \frac{2\pi}{k}$,

$$E(z) = E_0 e^{ikz} \left(1 + i\lambda L \sqrt{\mu_s} \right) \sim E_0 e^{i(kz + \lambda L \sqrt{\mu_s N})} \quad (2.10)$$

The second term of exponential can be considered as the phase of the incident wave, i.e.

$$\begin{aligned} \lambda L \sqrt{\mu_s N} &= \frac{2\pi}{\lambda} (n-1)L \\ (n-1) &= \frac{\lambda^2}{2\pi} \sqrt{\mu_s N} \end{aligned} \quad (2.11)$$

from Eq.2.7 and Eq.2.11 we have:

$$\begin{aligned} I_s &\sim \frac{1}{\lambda^4} \\ I_s &\sim \frac{1 + \cos^2(\theta)}{\lambda^4} \end{aligned} \quad (2.12)$$

where $\theta = 0$ is the forwarded scattering. The scattering intensity for different wavelength has been shown in Figure 2.6. In most of the tissues, both scattering and absorption occur simultaneously. The total attenuation is considered as the summation of the absorption and scattering coefficients.

Figure 2.6 Scattering intensity for different wavelengths [Vog03].

Figure 2.7 Five laser-tissue interaction regions [Kro89].

2.4 The Impact of Light Beam on Tissue

Since the advent of lasers, research into the interaction between biological tissue and laser light system types has never stopped. This interaction is now classified into five categories: (1) photochemical, (2) photothermal, (3) photoablation, (4) plasma-induced ablation, and (5) photodisruption. For any specific biological tissue ablation, laser parameters such as wavelength, pulse duration, and repetition rate; must all be factored into the surgeon's pre-procedure planning. Any medical procedure in the five categories shown in Figure 2.7 [Rob70] can be carried out successfully if an appropriate laser is chosen and the biological tissue interaction is estimated correctly. But, as can be seen in Figure 2.7 some of these interactions and mechanisms cannot always be accurately predicted. The best properties a laser can have for ablation procedures are mono-chromaticity and collimation. These two parameters give a necessary high-energy laser light in a narrow beam. These characteristics have been shown to remove tissue accurately and precisely [Sol61].

Laser ablation is a process where tissue or organs are removed by a high-power laser beam. The first medical application of a laser system was 1963, when a laser was used to repair a retinal tear in an ophthalmology surgery [Cam09]. In the early 1980s, a laser system was applied to orthopedic procedures. In early surgeries, continuous wave beam lasers with a long pulse were used for ablation. However, these laser systems caused carbonization, and tissue fracturing around the ablation sites

[Kro89; Nus88; Nie07]. Over time it was concluded that laser systems with higher irradiance had the most impact on tissue ablation surgeries. Laser irradiance is achieved by concentrating a low average energy laser beam spot onto a small biological tissue area. Such a laser system, when introduced by Myers [Ver12], had the ability to cut bone precisely with low carbonization and mechanical stress. Henn et al. [Hen12] then introduced CO₂ lasers into surgical procedures. The light intensity of these laser systems was magnified by using carbon dioxide molecules.

2.4.1 Photochemical Mechanism

A photochemical interaction uses a continuous wave laser light beam with a long pulse duration, low power density, and low-intensity light. Atoms and molecules excited by light energy react with their surrounding atoms and molecules and exchange electrons. This interaction produces stronger atoms and molecules than that held by the original atoms and molecules. Photochemical interaction generates radiation in the visible wavelength region. When a He-Ne laser with $\lambda = 635nm$ or a Rhodamine dye lasers with $\lambda = 632nm$ are used in a medical procedure, e.g., photodynamic therapy (PDT), a photosensitive agent is injected into the patient's veins and the laser's irradiation is applied after a wait period; i.e., allowing time for the chemical agent to destroy cancerous tumor cells. Fortunately, injected photosensitive agents disappear from healthy tissue after around 48 hours. But, their concentration does not decrease in tumor cells, particularly in cancerous cells that are rapidly proliferating and even after seven days. When the laser beam is applied to photosensitive agents, they interact with the radiation and destroy the tumor cells. Biostimulation and Photobiomodulation are two other mechanisms used in photochemical interaction. Again, these mechanisms occur at very low-levels of irradiation intensity, typically $1 - 5mW$. Many studies have been carried out into the chemical reaction of cells by absorbing light. These include, but are not limited to: (1) hair growth, (2) wound healing, (3) stimulated collagen synthesis, and (4) pain relief, etc. [Mes71]. For example, [Chu11] showed that cytochrome oxidase, which is an enzyme in the mitochondria, absorbs photons and either: (1) increases the adenosine triphosphate to enhance the cell's performance to remove infection, or (2) improves the reactive oxygen species (ROS) level to boost cellular repair.

Figure 2.8 The performance of different laser systems with the 3 watts output power on chicken liver [Nie07; Aok15].

2.4.2 Photothermal Interaction

In Photothermal interaction, irradiation energy is absorbed by tissue and converted into heat energy, which increases the temperature of the tissue. This particular mechanism causes vaporization, decomposition, and coagulation in the tissue, all of which depends on the maximum temperature of the tissue and its longevity. Both continuous wave and pulsed wave lasers have been applied to create the thermal interaction. For example, (1) an Argon laser with $\lambda = 514nm$ has been used for retina coagulation, (2) a near-infrared Nd:YAG laser with $\lambda = 1.06\mu m$ has been used to cut and coagulate soft tissue, and (3) an Er:YAG laser with $\lambda = 2.94\mu m$ has been used for cutting the soft and hard tissue. Figure 2.8 shows the performance of different lasers with the same output power (3 watts) in cutting and simultaneously coagulating chicken liver [Nie07].

2.4.3 Photoablation Interaction

A Photoablation mechanism is achieved using short wavelength, high power density lasers, e.g., between $10^7 - 10^{10} W/cm^2$. This mechanism breaks the molecular bond in biological tissue that leads to tissue fragmentation [Sri82]. The short wavelength, high power density lasers also reduce both the heat generated in tissue and the thermal damage to the surrounding tissue; mainly because short wavelength, high power density lasers, apply pulses with a short duration. Other high-energy

photon laser systems have also been applied in this type of biological tissue interaction, e.g., a laser system with UV spectral wavelengths. For example, excimer lasers have been applied to create a photothermal effect [Sri82].

2.4.4 Plasma-Induced Interaction

In this interaction, the intensity of the laser's energy is higher than the energy that binds electrons to ions ($\sim 10^9 W/cm^2$) and this leads to ionization and plasma formation. The photodisruption mechanism distributes energy through thermal diffusion. The biological tissue is heated past a temperature higher than a critical point a tissue phase transition occurs. Here, tissue transitions from a solid phase to a plasma phase, where the plasma absorbs all the available incident photons. This interaction has been applied in medical procedures such as lithotripsy, i.e., the removal of urinary stones. The excimer laser, Ho:YAG laser, and Ti:sapphire laser with high-energy and short pulses have also shown promising results when applied to urinary stone removal [Ten87; Ste89; Nie91].

2.5 Conclusion

This chapter reviewed literature related to the medical applications of laser systems. The range of laser systems in everyday use in medicine was reviewed, as were the interactions that take place between biological tissue and laser systems. This review began by considering the general light on surface interactions of scattering and absorption, with biological tissue being the surface in this instance. The review then considered more detailed light absorption issues, including: photo-chemical, photo-thermal, plasma-induced and photo-disruption interactions. The literature review concluded that prior to a medical practitioner performing a biological tissue ablation, they must choose a laser system with the appropriate wavelength, pulse duration and repetition rate. Prior to a performing a biological tissue ablation procedure the medical practitioner must optimizing these parameters in an attempt to ensure a successful patient outcome.

CHAPTER

3

ELECTRO-MAGNETICALLY OPERATED SENSORS, ACTUATORS, AND MICRO-ROBOTS, IN BIOMEDICAL APPLICATIONS

3.1 Introduction

Robot-assisted minimally invasive surgery is common in numerous medical fields because the procedure is carried out via small incisions and it is: (1) fast and stable, (2) precise, (3) has a reduced rate of infection, and (4) requires shorter hospitalization [Lan04]. Minimally invasive surgery technologies that are based on electro-magnetic fields are used widely in the designs of biomedical sensors and actuators. In part, this is because these sensors and actuators can be controlled wirelessly and they can be miniaturized. This chapter discusses how our research conducted into

total joint implant loosening detection and the design of a proximity electro-magnetic sensor, led to the development of a catheter actuator. Now, whereas if the electro-magnetic field current of a proximity sensors system is constant, the displacement of a permanent magnet within the field can be measured in the micron range. So, it was reasoned that the inverse could be applied, if the magnetic field current of an actuator system is varied, the displacement of a permanent magnet within the field can be displaced. The work in this chapter, along with the introduction of concepts from robotics and control, led to the design, fabrication and testing of the electro-magnetically controlled laser ablation system with feedback that is described in chapter 4.

3.2 Magnetically Actuated Microbots

Micro-scale robots follow the same fundamental laws as macro-scale robotic systems. However, for in vivo applications, because of their scale, they require precise design and construction to overcome their inherent design challenges of a surface to volume ratio, surface tension and viscosity (friction). Microbots are molecular devices that are used for medical applications such as the blood circulatory system. Recently microbots have been used in new medical applications, i.e., the central nervous system, the eye, and the urinary and prostate systems. Transferring the power to microbots remains the major challenge if they are to be used in both in vitro and in vivo biomedical applications. The use of on-board power supplies, energy harvesting, transferring power wirelessly, have all been applied to address the microbot power problem. Using an electro-magnetic field to control microbots wirelessly remains the most economical approach for now. In [Kha14] the motion of a self-propelled microbot was developed to operate inside a fluidic micro-channel under closed-loop electro-magnetic field control. Figure 3.1 shows the range of electro-magnetically actuated microbots that have been developed for use with stem cells, drug delivery, brachytherapy, and Hyperthermia. In the brachytherapy application, the electro-magnetically actuated microbots are used to transfer radioactive materials to a cancerous tumor, in order to remove the tumor by precisely placed radiated energy. Magnetically actuated microbots are also applied for thermally ablating cancerous tumors, by transferring precisely focusing thermal energy onto the targeted locations of unwanted cells. In drug delivery applications, microbots carry an appropriate drug under electro-magnetic actuation to a delivery site so as to decrease the side effects locally within the patient's

Figure 3.1 Different magnetically actuated microbots: a) Microbots applied in Brachytherapy to transfer the radioactive material to the targeted tumor [Dev07], b) An electromagnetic microbot actuator for drilling, steering and locomotion[Jeo11], c) An active electromagnetic based microbot to deliver drug to the interested region [Han16], d) A magnetic microbot system for human arrhythmias ablation [Cos16], e) A closed-loop magnetically actuated control of a microbot for removing blood clots [Kha19], f) A soft heating microbot system that uses thermosensitive magnetic materials to necrotize cancer cells. in[Sat02], g) A wireless magnetically actuated and self-expandable stent delivery system that can be used in narrow tubular cases[Lee17], h) Microbots applied in biopsy to sample the human tissue [Byu05].

body. Lastly, electro-magnetically actuated microbots have been used to remove tissue samples from inside the human body, i.e., a biopsy application, and successfully transferred the sample to the outside of the body for further analysis [Nel10].

3.3 Magnetically Actuated Wire-Guided and Catheter Microbots

In recent microbot research, permanent magnets have been integrated into soft materials. These soft-microbots are applied in biomedical applications to improve the control of the movement of microbots inside the human body. Soft-microbot motion control is once again using an externally applied electromagnetic field. A novel soft-microbot system was developed using Polydimethylsiloxane and controlling it with eight neodymium permanent magnets [Kim19]. Similarly, a ferromagnetic-based soft micro-robot was introduced by [Jeo19]. The soft-micro-robot in [Jeo19] was fabricated by combining a ferromagnetic material and an elastomer, i.e., NdFeB and thermoplastic polyurethane, respectively. A comparison between magnetically actuated guide-wire micro-robots and manually controlled guide-wire micro-robots is reported upon in [Kri06]. Experimental results show that the magnetically actuated guide-wire systems are faster, safer, and are very accurate. Examples of

Figure 3.2 Magnetically actuated guide-wire robots[Hwa20]: a) A magnetically actuated guidewire system with additional gradient coils [Lal15], b) A magnetic navigation system (MNS) that controls the deflection of a microguidewire by the generated magnetic field[Kri06], c) A soft microbot system controlled by external eight electromagnetic coils [Jeo19], d) Schematic of a flexible microrobot that is controlled by the electromagnetic field to improve guidewire steerability [Jeo18], e) Schematic of the magnetic guidewire with hydrophilic coating[Tsu06], f) soft robot is fabricated by combining ferromagnetic material such as NdFeB and an elastomer like thermoplastic polyurethane[Kim19].

magnetically actuated guide-wire robots are shown in Figure 3.2 controlled catheters are mainly applied in cardiac ablation, to remove tissue that causes an abnormal heart rhythm called atrial fibrillation (AFib). In a similar manner to guide-wire micro-bots, the motion of a magnetically actuated catheter is controlled by an externally applied electro-magnetic field, which navigates the catheter through the veins and reach the desired target site on the heart. Figure 3.3 depicts different developed magnetically actuated catheters that can be applied in biomedical applications. Figure 3.3.(a) , shows a variable stiffness (VS) device, one where the tip of the catheter is controlled precisely through the vein via an externally applied electro-magnetic field [Cha17c; Cha19]. The stiffness characteristics of the tip of this catheter are altered by separately applying thermal energy to it, i.e., by applying heat. In [Fad02] the authors compare the performance of a magnetically actuated catheter and a manually controlled catheter. The results showed that the magnetically actuated catheter was faster and more accurate [Fad02].

Figure 3.3 Different developed magnetically actuated catheters that can be applied in biomedical applications[Hwa20]: a) A variable stiffness device (VS) in which the tip is controlled precisely through the vein via an external electromagnetic field [Cha19], b) Semi-Automatic magnetically actuated method to control and direct an intravascular three-magnet tip in human heart[Le16], c) A magnetic remote navigation system that controls the catheter in in real-time[Gan11], d) A magnetic guidance system (MGS) that controls an ablation catheter through the generated electromagnetic field [Fad02], e) Schematic and the manufactured tethered magnet designed for 5-DOF positioning for cardiac ablation[Cha17b], f) Different design of ferromagnetic tips and a system with a high gradient coils to maximize the deflection [Gos11].

3.4 Electromagnetically-Based Sensor in Biomedical Applications

The electromagnetic-based Medical devices are generally driven by permanent magnets, electromagnetic coils, or both of these components. Mainly the permanent magnets based devices are utilized to screen and diagnose the disease in the interesting area in biomedical application. Dominguez in [Dom09] presented a technique that uses permanent magnets graspers for retraction and triangulation to create a sufficient and safe surgery field for minimally invasive surgery. A magnet camera system was developed by [Ter12] for laparoscopic surgery, which does not interfere with the surgical tools and provides enhanced view during the operation. A compliant capsule endoscopic system was designed by [Yim11] in which the movement of the capsule is controlled by the external electromagnetic field generated by the permanent magnets. Figure 3.4 shows several developed systems that are driven by permanent magnets and are used in biomedical applications. The medical devices, which are controlled and driven by the electromagnetic field generated by the coils, are mainly larger than permanent magnet systems and are used for endoscopic navigation systems, vascular and Gastrointestinal tract. Figure 3.5 depicts some of the electromagnetic based sensors applied for positioning and monitoring in biomedical applications. A toroidal electromagnet system

Figure 3.4 Several developed systems that are driven by permanent magnets and used in biomedical applications [Sli15]: a) Permanent magnets-based graspers used for safe minimally invasive surgery [Dom09], b) A magnet camera system for laparoscopic surgery [Ter12] c) A compliant capsule endoscopic system controlled by external magnetic field [Yim11], d) An in vivo insufflation capsule that its valve is controlled by an external electromagnetic field [Pas14], e) A Magnetic sphincter applied in Gastroesophageal reflux disease (GERD) [Gan08], f) A Local Magnetic Actuation to control and tilt tuning of a Laparoscopic camera [Sim12], g) A Magnetic air capsule robotic system with an endoscopic grasper [Val11].

is presented in [Luc15] (Figure 3.5.(a)) in order to control and drive a magnetic endoscopic capsules. The proposed technique by [Dil13] (Figure 3.5.(b)) is able to independently handle the movement of multiple microbots in a 3D environment using magnetic gradient pulling.

3.5 Design of an Electromagnetic-Based Wireless Sensor System for Detecting Loosening in Total Hip Replacement

Total hip replacement (THR) is a surgical procedure that is used to relieve pain while retaining hip functionality. THR was first introduced by Charnley in early 1960 [Jac12]. It is estimated there are over seven million people in the United States living with a total joint replacement [MK15]. It is projected that by 2030, the annual number of joint revision surgeries will range between 102,296 [Slo18] and 364,847 [Kur07] at a cost to the system between 7 and 25 billion dollars annually. Today, THR is a successful surgical procedure, with a success rate up to 95%. However, the implants can fail, at the rate of 5%. According to Bayliss et al. [Bay17], for patients aged 70 years at implantation, the lifetime risk of revision (LRR) surgery has been shown to be between 4.4% and 7.7%. For patients aged between 60 and 70 years at the time of primary surgery, LRR increased with decreasing age,

Figure 3.5 Electromagnetic based sensors and actuators: a) A toroidal electromagnet to control to a magnetic endoscopic capsule [Luc15], b) Magnetic gradient pulling to control multi microbots [Dil13], c) A magnetic shielding technique to control the generated force by the external permanent magnet applied Minimally-Invasive procedures [Bre08], d) A wireless control navigation system that controls magnetic microbots by generating the tailored magnetic fields [Luc14].

reaching approximately 15% for both hip and knee replacement at 60 years, with greater risk in men. However, a significant increase in LTRR has been seen in relatively younger men, with values up to 29.6% for knee replacements in patients between 50–54 years.

The underlining reason for failure of a total joint replacement is component loosening, be it from infection, aseptic loosening, or wear and tear. The earlier pathologic loosening is detected, the greater are the chances for successful remedial measures, including revision surgery. Unfortunately, the present diagnostic radiographic tools used to determine component loosening are imprecise. The resolution of standard radiographic imaging is in the millimeter range and is insufficient to detect early stages of implant degradation or loosening which occur in the micron range. So, the early detection of THR loosening will help surgeons address the problem before patient pain and discomfort increases, and patient mobility decreases. For the medical practitioner, the early detection of hip joint loosening assists in planning remedial surgical intervention, thereby reducing the time and cost of surgery. As a prosthesis loosens, the surrounding supporting bone erodes. The longer this process continues, the less likely are the chances for a successful revision surgery. Based on a comprehensive literature search, early hip prosthesis loosening is detected by: (1) medical imaging, or (2) non-medical imaging, e.g., accelerometer vibration. Currently the favored diagnosis tool for determining prosthesis loosening is medical imaging; even although such techniques do not provide the practicing orthopedic surgeon with objective data [Pal14; Nil94; O84; Oye96; Pfa98; Tem05; Rut12; Ara17]. The two most common imaging techniques are: (1) computed tomography (CT), and (2) molecular imaging, e.g., positron emission computed tomography (PET). The major

Figure 3.6 General view of the Hip Loosening Detection Using Inductive Proximity sensor.

disadvantages of these methods are: (1) the amount of the radiation exposure to the patient, and (2) the high cost of testing [Koo19]. Most non-imaging techniques used to detect hip joint loosening use sensors that detect mechanical vibration [Ros89; Gla04; Geo01; Als17]. So these techniques are based on the response of a detected output signal from the joint prosthesis to an applied input signal.

The most commonly used method for detecting hip implant loosening is frequency response, where detection is derived by: (1) studying the input/output signal relationship over a range of frequencies, and (2) spectral analysis, i.e., analyzing the output signal through decomposing it into a number of discrete frequencies. In experiments the frequency response of a secure hip (no loosening) is compared with the frequency response of a suspected loosened hip. Loosening is based on a study of the higher order harmonics, if higher order harmonics have an output signal amplitude 25% greater than the fundamental frequency amplitude then the hip implant is loose[Rie13]. However, this technique is not sufficiently accurate to detect early hip loosening. Telemetric approaches have been used to detect hip prosthesis loosening. Here acceleration sensors capture an induced mechanical vibration signal and transmit the recorded data outside the body for further analysis. However, none of these non-imaging methods provide information about the amount of loosening as a metric[Pue00; Mar09; Sau12; Su17; Su15]. Non-imaging Electromagnetic (EM) research helps surgeons diagnose diseases, e.g., Lambert and Gremels in (1926), measured the electrical resistance across lungs and to monitor for pulmonary edema [Lam26]. In early 1936 the first EM flow-meter was

Figure 3.7 LC tank circuit. a) Series model. b) Parallel Model.

created to measure blood-flow [Kol36; Wil71]. In 1971 microwave radiometry devices were applied to detect lung diseases [Ped78]. This section focuses on the design, development, and application of an EM-based inductive proximity sensor system for detecting hip joint implant loosening in the micron range [Moh20]. A new inductive proximity sensor system was designed and fabricated to measure the level of hip loosening, see Figure 3.6. As can be seen in Figure 3.6 the loosening of the implant is detected by analyzing the EM field generated by the inductive proximity sensors located around the implant. The EM field induces an eddy current on the surface of the implant. The circulated current generates another EM field in the opposite direction of the original field and changes the parallel resistance, inductance and, resonance frequency of the sensor. Variation of these parameters indicates positional change of the implant. The loosening of the implant can be accurately measured from the sensor's quality factor (Q), series resistance, and the inductance.

3.5.1 Theoretical Modeling of Implanted Proximity Sensor

Inductive-capacitive (LC) passive sensors were first used by Collins to measure pressure in the eye in 1967 [Col67]. Due to the advances in micro-electro-mechanical systems (MEMS), and the reduction in scale that this technology brings, LC circuits are now used widely in research related to biomedical engineering systems and bioengineering. One of the main advantages of the passive LC circuits is their ability to obtain sensed data without physical contact. The orthopaedic surgeon that initiated this project has performed well over 1000 total hip replacement surgeries. That surgeon is confident that the in vitro modeling of the sensor experiment closely approximates to the in vivo

conditions that are typically encountered. The sizing of the coil was modeled to configure with the typical diameter of the intramedullary canal of the human femur. The width of the femoral model corresponds to the typical width of the human femoral cortex. The Coils placed in other areas around the prosthesis can be made smaller than the typical 10 mm intramedullary coil since they may be placed closer to the target, i.e., in the region of 3 mm to 5 mm . Figure 3.7 shows a simplified circuit diagram of an LC proximity sensor system. In operation, the sensor system is designed to oscillate at a specific frequency and losses appear as a series or parallel model. By moving the conductor target, i.e., the resonant frequency, the magnitude of both the current and the voltage of the sensor are altered. Under these conditions the system increases the power needed to drive the sensor system to maintain the desired oscillation amplitude, that is tracked and servo-controlled to a set point. The sensor system's parameters are a function of the difference between the ideal (without implant) and the actual (with implant) system input power. The sensor detects the motion of a metal implant with respect to a sensor location. The induced eddy currents are directly related to the position, size and the material used for the implant. By moving the metal implant in relation to the coil, energy is transferred to the implant and an eddy current is generated. The induced eddy current generates its own EM field, which is a reaction to, and in the opposite direction to, the original electromagnetic field. This reaction can be considered as a coupled inductors. The primary winding is the sensor inductor and the secondary winding is the eddy current. The coupling between the winding changes the general impedance, inductance, Quality factor (Q) of the sensor system. The quality factor (Q) of the coil, the inductance and parallel resistance, are then directly related to the distance to the metal implant. In a parallel LC tank system, if the main loss in the circuit is due to the resistance of the coil, the quality factor (Q) and the resonance frequency of the tank can be calculated using Eq.(3.1) and Eq.(3.2), see [Bil10]

$$Q = \frac{\text{Stored Energy}}{\text{Average Power dissipated}} = \frac{1}{R_s} \sqrt{\frac{L}{C}} = \frac{\omega_0 L_s}{R_s} \quad (3.1)$$

$$f_0 = \frac{1}{2\pi\sqrt{LC}} \quad (3.2)$$

where, in Figure 3.7.(a), L and R_s are the inductance and AC series resistance of the coil. C is the fixed capacitive sensor in parallel to the coil inductance.

Various coil geometries can be applied for designing the proximity sensor. The most commonly applied proximity sensor geometries are based on flat spiral coils, or a solenoid.

In this sensor system design the skin effect causes the effective series resistance of the inductor to increase at higher frequencies. At the higher frequencies, when the skin depth is small, the effective cross-sectional area of the conductor is reduced. The resistance of the wire increases and as a result the series resistance in the LC tank increases. The series resistance can be derived by considering the skin and proximity effect as in Eq.(3.3) and Eq.(3.4) [Kni13],

$$R_s = \frac{\rho l}{A_{eff}} \quad (3.3)$$

$$A_{eff} = \delta \pi d \quad (3.4)$$

$$\delta = \sqrt{\frac{\rho}{\pi f \mu}} \quad (3.5)$$

where ρ is the resistivity of conductor ($\Omega.m$), l is the length of conductor, A_{eff} is the effective cross section area of conductor, δ is the skin depth, f is the operation frequency, and μ is the magnetic permeability of conductor. The effective area in Eq.(3.4) is valid when the skin depth is between zero and the wire radius (r). If the skin depth is larger than the wire radius, then the AC resistance of the wire will be the same as DC resistance ($R_{AC} = R_{DC}$). For detecting loosening, since the sensor is operating at a higher frequency, the skin effect should be taken into consideration; because it changes when the implant moves in relation to the sensor. Also, the series resistance (R_s) of the inductor can be modeled as a parallel resistance, see Eq.(3.6)

$$R_p = \frac{L}{R_s C} \quad (3.6)$$

The parallel resistance (R_p) is an important parameter in the measurement of target movement. Because the value of the parallel resistance is high compare to the series resistance it is best measured using an instrument with high resolution. The following readings must be measured and recorded, inductance, parallel resistance, and the Q factor of the coil when a metal target is placed close to the LC tank. In order to increase the sensitivity of target movement detection the flowing parameters

must change: (1) increase the target area; for increased magnetic flux, (2) increase the magnetic field, and (3) increase the cross section area of the sensor face. Essentially, to detect implant loosening a sensor with a high Q factor is desirable. Note that based on (3.1), increasing the sensor current will not help to increase the sensing distance because increasing the current of the coil will increase the stored energy as well as dissipated energy. As a result the Q factor will remain constant. In order to mathematically model the inductive proximity sensor Maxwell's equations are used, Eq.(3.7)

$$\int_S \nabla \times b \, da = \oint_C b \cdot l \, dl = \mu_0(I_{enc} + \epsilon_0 \frac{d}{dt} \int_S e \cdot \hat{n} \, da) \quad (3.7)$$

where b is the magnetic flux, e is the electric field, I_{enc} is the enclosed current, ϵ_0 is the electric permittivity of free space, and μ_0 is the magnetic permeability of the free space. Also, Maxwell's equations can be expressed in frequency domain,

$$\nabla \times H = J \quad (3.8)$$

$$B = \nabla \times A, \quad \text{and} \quad B = \mu_0 \mu_r H \quad (3.9)$$

$$E = -j\omega A \quad (3.10)$$

$$J = \sigma E + i\omega D \quad (3.11)$$

where H and E are magnetic fields electrical fields respectively, B is magnetic flux density, D is displacement current density, σ is electrical conductivity, μ_0 is magnetic permeability in free space, μ_r is the permeability of the material, ϵ is dielectric permittivity, and ω is the operating frequency, see [Tur13].

3.5.1.1 Modeling and Simulation of The Proximity Sensor

In this section the design, modeling and simulation of LC tank are carried out using COMSOL software and finite element analysis (FEA)[COM18]. The module "Magnetic Field" was used to

Figure 3.8 3D Proximity sensor simulation a) Solenoid Coil b) Circular Flat spiral coil. An electrical current of $25mA$ was applied to the coils which generated the magnetic field (blue lines) shown and interacting with the metal implant.

calculate the properties of the coils and the EM fields. The value of parallel capacitor was tuned to satisfy the desired resonance frequency, Eq.(3.2). A three-dimensional model was constructed in COMSOL for two coil designs and is shown in Figure 3.8: (1) a 4 layers circular flat spiral coil, simulating a coil on a printed circuit board, and (2) a wire-wrapped solenoid coil with 45 layers using 35 AWG gauge insulated magnet wire. The model also includes a chromium-cobalt metal rod with $9mm$ diameter to simulate the femoral component of a metal hip implant. In order to optimize the accuracy of hip loosening detection at increased implant distances, two different geometries (flat spiral coil and solenoid coil) were analyzed at different resonance frequencies and implant displacements. Eqs.(3.1)-(3.11) are used by COMSOL to model the proximity sensor, EM field, skin effect, coupling between windings and series resistance.

3.5.1.2 Effect of Variations in the Solenoid Coil Geometry

The geometry of solenoid coil and its operating frequency play an important role in determining the:(1) amount of energy dissipated by the implant, (2) eddy current generated on the surface of the implant and, (3) size of the sensor. In order to optimize the implanted sensor design, to ensure it is capable of loosening detection, the solenoid coil was modeled and analyzed for a variety of geometries. Table 3.1 and Table 3.2 summarizes the simulation results for solenoid coil with different scenarios which are derived from the results in Figure 3.9. Simulations were run using COMSOL finite element analysis, with air as the medium. The Co-Cr rod (ASTM 1537 material) was inserted

Figure 3.9 Different Coil geometries. a) Single layer coil with 45 turns without ferrite core. b) single Layer coil with 45 turns and rod ferrite core. c) Two layers coil with 15.5 turns each layer and rod ferrite core. d) Two Layers coil with 15.5 turns each layer without ferrite core.

at 11 mm distance to the coil and represents the implant. In some configurations a ferrite core was placed inside the solenoid coil to increase the magnetic flux. A ferrite material with a relative permeability of 2000 was used (ferrite 77 material). Power loss in a metal object is represented as the surface current density (A/m) and volume loss density (W/m^3). Also, 35 AWG wire was used to simulate the 1-layer and 2-Layers solenoid coils. In 2-layer solenoid coils the solenoids are wound with two isolated layers. The inner layer and the outer layer can have different number of turns. The same number of turns were modeled for the 2-Layers solenoid coils. The six solenoid coils modeled and simulated had different: (1) numbers of turns, (2) numbers of layers, (3) ferrite cores, and (4) no ferrite cores. Electrical currents, at desired frequencies, were applied to the coils using COMSOL software. The resulting Q as well as induced current and power dissipation in the Co-Cr rod were calculated by the software. As can be seen from the results, (Figure 3.9, Table 3.1 and Table

Table 3.1 Solenoid Coil Geometries Used for COMSOL Simulations.

	Coil A	Coil B	Coil C	Coil D	Coil E	Coil F
Coil dia. (mm)	10	10	10	10	10	10
Coil Length (mm)	13	13	13	13	6.5	6.5
# turns / # Layers	45 / 1	45 / 1	45 / 1	45 / 1	15.5 / 2	15.5 / 2
Ferrite rod (6 mm dia.)	no	yes	no	yes	no	yes
L (μ H)	10.88	28.28	10.88	28.28	8.44	28.14
Tank capacitor (pF)	2018	776	644	248	831	249
Frequency (kHz)	1074	1074	1900	1900	1900	1900

Table 3.2 Simulated Results for Solenoid Coils derived from Figure 3.9.

	Coil A	Coil B	Coil C	Coil D	Coil E	Coil F
Q, baseline (no implant)	99	254	175	440	175	439
Q, 11 mm distance	97	238	172	401	172	404
% change in Q	2.0%	6.3%	1.7%	8.9%	1.7%	9.2%
Surface current ($\times 10^5$ A/ m^2)	1.0	2.6	1.5	3.9	1.4	3.7
Volume loss density ($\times 10^4$ W/ m^3)	0.9	5.9	2.0	13.3	1.8	12.1

3.2) coils with ferrite core induced increased magnetic energy on the Co-Cr rod, and had greater sensitivity (decrease in Q with rod placement at 11 mm), compared to coils with no ferrite. Also, full-length and half-length coils showed a comparable induced magnetic energy and Q sensitivity values. By increasing the frequency the Q factor of the proximity sensor will be changed. Based on the results, at 1900 kHz resonance frequency the coils had higher Q and induced greater current/power dissipation compared to 1074 kHz coils, even with similar inductance values. Power dissipation is based on the surface current and volume loss density of the implant. Based on the results recorded in Table 3.1 and Table 3.2, for the sensors with 1900 kHz frequency there is a greater power loss and surface current in comparison to sensor systems at 1074 kHz. The Q factor and power dissipation indicates increased sensitivity in the 1900 kHz coils due to an increased coupling with the Co-Cr rod.

3.5.1.3 Effect of Blood and Bone On Hip Loosening Detection

In this section, the proximity sensor was tested for conditions where different mediums existed between the implant and the proximity sensor, i.e., air and biological material; such as blood and bone. Three LC tanks with three different resonance frequencies and the same Q factor were modeled with biological material as the medium. Since the LC tanks have the same Q factor, the coil properties

Figure 3.10 3D model of the proximity sensor with cortical bone and blood material. Different dielectric properties, from Table 3.4, are considered to model bone and blood tissues in COMSOL software.

such as length and number of turns required to be adjusted to comply with this constraint. Table 3.3 presents the properties of the modeled LC tanks. In this case Q_{max} is 57.3 for each tank. The 3D model of the sensor was shown in Figure 3.10. Based on Figure 3.10 cortical bone (femur) section was added around the sensor coil (30 mm outside diameter, 10 mm inside diameter), the media between the sensor and Co-Cr rod was considered to be blood, and the boundary condition (sphere with 100 mm diameter) around the construct was modeled as blood. The conductivity, permeability and permittivity of the cortical bone and blood at different frequencies are summarized in Table 3.4[Gab96]. Simulation results related to implant displacement are shown in Figure 3.11.(a). The simulations were run for a chosen set of resonance frequencies. As can be seen from the results the Q factor of the LC tanks increase as the implant displacement increases, and eventually the curves flatten out and reach the same maximum Q factor. At the lower frequencies (106 kHz) the LC tanks are more sensitive. Figure 3.11.(b) shows that irrespective of whether the sensor coils

Table 3.3 Three LC tanks specification at resonance frequencies 1 MHz, 265 kHz and 106 kHz respectively.

	Specification of the coil	Inductance	Capacitor	Resonance Frequency
Tank A	1 layer, 45 turns (35 AWG wire)	$L = 13.6\mu\text{H}$	$C = 1.86\text{nF}$	1MHz
Tank B	5 layers, 38 turns (35 AWG wire)	$L = 200\mu\text{H}$	$C = 1.80\text{nF}$	265kHz
Tank C	5 layers, 40 turns (35 AWG wire)	$L = 200\mu\text{H}$	$C = 11.25\text{nF}$	106kHz

Figure 3.11 Q Factor vs. Implant displacement for the tanks with the same Q factor and different resonance frequencies in the material representing biological properties. a) Q factor of the LC tanks vs. displacement. b) Q factor of LC tanks vs. displacement for medium material as blood and cortical bone.

Table 3.4 The dielectric properties of cortical bone and blood.

	Conductivity	Permeability	Permittivity
Cortical Bone @ 106 kHz	2.08E-2	1	2.25E+2
Cortical Bone @ 265 kHz	2.13E-2	1	1.95E+2
Cortical Bone @ 1 MHz	2.44E-2	1	1.45E+2
Blood @ 106kHz	7.03E-1	1	5.11E+3
Blood @ 265kHz	7.17E-1	1	4.78E+3
Blood @ 1MHz	8.22E-1	1	3.03E+3

were embedded in either air, or bone and blood, i.e., representing biological materials, they gave similar results. The sensitivity remains relatively unchanged when the sensor is simulated in a bone and blood environment. Similar results were obtained with the physical experiments, where a conductive saline solution was used to emulate bone and blood.

3.5.2 Experimental Results

This section records the results obtained from a physical proximity sensor fabricated and tested by RTI company under the same conditions used in modeling and simulation trials, see Figure 3.12. The sensor coil was mounted on a PC board and the wire leads from coil were soldered to PCB traces leading to an SMA (coax) connector. The parallel tank capacitor was mounted on the underside of PCB, in parallel with the inductor coil. The SMA cable was attached to an LDC 1101 board, which

Figure 3.12 The experimental test-bed to measure loosening of hip joint implant.

measured the proximity sensor parameters, i.e., inductance and parallel resistor readings. The Co-Cr rod (ASTM 1537) was used to model the actual implant and was mounted to a Thor Labs translation stage. The translation stage was actuated by a manual linear actuator, a micrometer, with 0.001" (25 μ m) resolution accuracy.

3.5.2.1 Test and Analysis of the Designed Proximity Sensor

A 10mm Co-Cr rod used as a proximity model for the tip of the distal prosthesis was, in the opinion of the orthopaedic surgeon, a reasonable model for present day hip prostheses. Other metal composites, such as titanium, will also need to be tested but, it is anticipated that any conductive alloy will can be tested using this approach. Under the same conditions as used in the modeling and simulation trials measurements were recorded from a 10mm solenoid sensor coil as a function of distance to the Co-Cr rod. In addition, the sensor was analyzed with and without a ferrite rod (6.3mm diameter) core, and in air and phosphate buffered saline (PBS, to replicate blood) mediums. In the laboratory tests, where the saline solution surrounded tubular solenoid coils (inside diameter 20mm) the Q factor was not impacted by immersion in saline. With the tubular solenoid coils the results gave equal resolution whether tested in air or in the saline solution. The LC tank was tested at 1074 kHz frequency and the values of the inductors and parallel capacitors are tuned to satisfy the resonance frequency. The results recorded from the physical experiment is cataloged in Figure 3.13. Based on the results, the Y axis of Figure 3.13 represents the digital outputs of LDC1101 device which

Figure 3.13 Sensor system impedance (digital output of LDC1101) vs. implant displacement for different case studies with coils tested at 1074kHz.

are corresponding to the impedance of the sensor. By increasing the implant distance the coil parallel resistance (digital value) will increase. Also, the sensor was tested under 1900kHz resonance frequency for different implant displacements. The results are shown in Figure 3.14 which are for narrower displacement ranges. The results show that the displacements can be resolved in a static measurement scenario in saline, i.e. impedance changes for fixed distances between the rod and sensor system. The digital measured data changes in regard to whether the sensor coil is tested in either air or in a saline solution. Also, the measurements were not affected by a saline solution fill as the medium.

3.5.2.2 Analysis of the Implant Movement with Continuous Oscillation

In order to test the sensor system under continuous movement, an LC tank with a 1900 kHz resonance frequency was considered. By applying a continuous sinusoidal oscillation to the implant and analyzing the output response the resolution of the sensor system was verified. The Co-Cr rod was placed at a fixed distance from the sensor coil and then oscillated at 0.7 Hz. This was achieved using the Vernier mounted on the translation stage, and tested at two different amplitudes 50μm (0.002") and 100μm (0.004"). Then, Fast Fourier Transform (FFT) functions were applied to convert the time domain signals obtained to a frequency domain spectrum. The results are shown in Figures 3.15-3.17. All tests were conducted with a saline solution inside the acrylic tube, Figure 3.15. As can be seen from the recorded measurements the FFT filter produces frequency domain plots which

Figure 3.14 The experimental results for implant displacement. a) Displacement from 3mm-3.3mm. b) Displacement from 5.05mm-5.25mm. c) Displacement from 7mm-7.15mm. d) Displacement from 8.25mm-8.4mm. The coils were tested at 1900kHz.

represent motion at two different amplitudes, i.e. $50\mu\text{m}$ and $100\mu\text{m}$. The FFT approach was used in order to analyze the proximity sensor's performance under large displacements. The FFT analysis was adopted because it was considered that the presence of measurement noise along with large implant displacements would be complex to recognize good resolution in the time domain. By analyzing the sensors recorded signals in the frequency domain the system is capable of increased measurement accuracy for large implant displacements. As can be seen from Figures 3.16-3.17, for the system system tested in Figure 3.15, $100\mu\text{m}$ resolution can be achieved up to distance of 20mm. Based on the results, the amplitude of the FFT signal is an indication of strength of the oscillation for a given oscillation amplitude ($100\mu\text{m}$ or $50\mu\text{m}$ oscillation). As a result, the two peaks (100 or $50\mu\text{m}$) can be delineated up to 12.7mm would indicate that the system is able to resolve $50\mu\text{m}$ up to that distance. At 15.2mm to 20.3mm distance, $100\mu\text{m}$ peak can be resolved, but not the $50\mu\text{m}$ peak separately. This would indicate that resolution is $100\mu\text{m}$ at these distances.

Figure 3.15 Experimental testbed for continuous oscillation. Two different scenarios are considered to verify the experimental testbed. Right: movement of the implant in horizontal direction, Left: Continuous Oscillation movement of implant.

3.6 Conclusion

In summary, this chapter dealt with the design, modeling, and physical testing of a new wireless inductive proximity sensor system for detecting hip joint implant loosening were presented. By moving a conductive object, such as a hip implant relative to a coil, the loss of energy due to the eddy currents in the hip implant alters the LC tank series and parallel resistances of the coil. These resistance changes are directly related to the distance between the object and the coil. First, the inductive proximity sensor was modeled and analyzed using COMSOL to optimize the sensors coil design, i.e., to increase the sensors sensitivity and resolution. Numerous coil configurations were considered, including the use of a solenoid coil. Results showed that when a solenoid coil was incorporated into the sensor system design, the resolution improved, and implant prosthesis loosening was readily detected. The results also showed that a ferrite core increases sensor resolution and the range over which loosening can be detected. In both simulated and physical experimentation, all sensor designs were analyzed in the two possible mediums that can exist between a sensor coil and a prosthetic implant, i.e., air and the biological material (blood and bone). Simulation results showed that the biological material had little impact on the resolution of the sensor. When the physical sensor system was fabricated and tested, the experimental results showed that the proximity sensor was capable of detecting movement, with high resolution, over a reasonable range of distances. In order to detect the implant loosening at higher distances with high accuracy, sensor data was analyzed in the frequency domain using FFT analysis. The results showed that by analyzing the signals in this way, loosening was detected at 20mm with $100\mu\text{m}$ resolution and that physical experimental results

Figure 3.16 Raw signals measured by the LDC 1101 for oscillation of the Co-Cr rod at 0.005mm and 0.1mm amplitude as well as raw noise signals (no oscillation). Each graph represents a single average distance between the solenoid coils to Co-Cr rod ranging from 2.54mm to 20.32mm . Solenoid coil operating at 1900kHz in saline solution.

Figure 3.17 Frequency domain signals calculated by fast Fourier transform of the raw signals (Figure 3.16) for oscillation of Co-Cr rod at 0.05mm and 0.1mm amplitudes. Each graph represents a single average distance between the solenoid coil to Co-Cr rod ranging from 2.54mm to 20.32mm . Solenoid coil operating at 1900kHz in saline solution.

were a good match to the simulated results. Through carrying out this research and by studying the European Union μ Ralp research project into developing new laser-based endoscopic tools for tissue ablation surgery [Ace17], it became apparent that this research in this chapter could be adapted to the laser ablation task. By varying the current being applied individually to four solenoid coils and attaching permanent magnets to a laser fiber optic cable, the movement of the fiber optic could be accurately controlled. And, if a photo-detector sensor was placed to measure the reflected visible beam, closed-loop feedback control could result. No other such laser research test-bed in biomedical tissue ablations, commercial or otherwise, appeared to possess closed-loop feedback control of laser light beam.

CHAPTER

4

A CLOSED-LOOP ELECTRO-MAGNETIC LASER SYSTEM DESIGN (PROTOTYPE 1)

4.1 Introduction

This chapter presents the design and development of an electro-magnetic laser scanning system, a new endoscopic tool, for performing biological tissue ablation in microsurgery. An electro-magnetic actuation system was designed and fabricated to control the movement of the tip of the fiber optic cable. First, minimally invasive laser ablation systems are introduced. These are laser systems where the positional control of the fiber optic cable tip is determined by a torque generated in a contactless Electro-Magnetic field. In the prototype 1 system under discussion, tip positional control was achieved by sending four current values, computed by the systems controller, to the four individually actuated solenoid coils that are the fiber optic cables actuator system. The interaction between the Electro-Magnetic (EM) field and the miniature permanent magnets hosted by the fiber optic cable, in a N-S-E-W configuration, generates the torque that produces a 2D scanning system.

Figure 4.1 General view of the closed-loop magnetically actuated laser scanner system.

According to an extensive literature search prototype 1 appears to be the first laser scanner system to use closed-loop control, i.e., by feeding-back photo-detector measurements to the controller. Within the prototype 1 endoscopic tool two plano-convex lenses are used to collimate the laser beam, i.e., to increase the laser systems focal length. This design alteration within prototype 1 increases the distance from the tip of the fiber optic cable to the target, i.e., the biological tissue being ablated. The thermal and structural deformation of the lenses were modeled, using the COMSOL software, to ensure their integrity was maintained when operating. The details of this new laser scanner system design and fabrication are outlined in the remainder of the chapter.

4.2 The Prototype 1 System Architecture

The closed-loop magnetic laser scanner system described in this sub-section was developed to improve the positional control accuracy of an endoscopic tool, by accurately controlling the positioning of a laser light beam emitting from the tip of a fiber optical cable. Figures 4.1-4.2 show the general architecture and layout of the closed-loop magnetic scanner system that became known as prototype 1. The prototype 1 endoscopic tool system consists of: (1) a flexible fiber optical cable, which can either carry a powerful continuous wave laser beam or a pulsed wave laser beam to the distal side of the endoscopic tool, (2) four solenoid coils, (3) five permanent magnets, (4) two

Figure 4.2 General view of the closed-loop laser ablation system.

plano-convex lenses, both to magnify the laser beam and to increase the distance to the biological tissue site, (5) a beam splitter which reflects the visible light, and (6) a photo-detector sensor to measure the actual position of the laser light beam on the biological tissue. The data gathered by the photo-detector sensor is fed-back to the systems controller, to reduce the positional tracking error, i.e., by becoming a closed-loop control system. The laser light system produces two types of laser beam: (1) a mid-infrared (IR) light, and (2) a visible light, for sighting in and calibrating the laser light beam positional control system. Both of these light sources are carried by a single fiber optic cable. Due to the material properties of the beam splitter, experiments showed that 90% of visible light was reflected to the photo-detector sensor. In a similar experiment with an IR light beam the beam splitter transmitted 90% of IR beam to the biological tissue site. So, a 90% reflection rate from visible light was used for sighting in and calibrating the positional control system, i.e., for closed-loop feedback control.

4.3 Prototype 1 Design

In order to design an automated magnetic laser scanner system a number of technical constraints were considered. First, the maximum strength of the endoscopic tool actuators system was designed, i.e., the maximum strength of the electro-magnetic field of the actuator system, and its resulting torque, needed to be estimated. This power calculation is required at this initial stage, because this

actuator system controls the position of the fiber optical cable, i.e., it is the main positional control system for the prototype 1 endoscopic tool. COMSOL modeling software was used to calculate both the actuator torque and to simulate the desired electro-magnetic fields of the system. The second major constraint considered when designing the prototype 1 system, was the dimension of the cylindrical housing that was required to carry the fiber optic cable, beam splitter, and lenses. Initially, a cylinder of 30mm diameter was estimated. However, after a design review the diameter of the cylinder housing was reduced to less than 10mm.

The cylindrical housing of the prototype 1 endoscopic tool was designed using Fusion 3D software. This cylindrical housing held: (1) four solenoid coils, (2) an optical fiber cable, and (3) two collimated lenses. The four solenoid coils were designed as 150 turns of 2.5mm diameter copper wire and a final length of 6.0mm. The two optical lenses were again designed using COMSOL software. These two lenses were required to handle the energy generated by the 20 watts holmium laser being considered for use with the lenses. This required that the lenses had to cope with a laser wavelength of 2100nm and a focal length that can be adjusted to 150mm. The optical fiber cable was fixed at one of the cylindrical housing. Permanent magnets with an outer diameter of 1mm and an inner diameter of 0.5mm were attached around the fiber optical cable. These permanent magnets operated under the control of the electro-magnetic field generated by the solenoid coils. The beam splitter was placed behind the optical lenses, in order to transmit the visible light to the target and to deflect light to the photo detector sensor. As stated in sub-section 4.2 the beam splitter transmitted 90% of an 2100nm IR wavelength, and 10% of a visible light. The photo-detector sensor was positioned to detect the reflected visible light from the beam splitter and measured the position of the light based on its intensity. The measured position of the laser beam was converted into a

Table 4.1 The components used for prototype 1 of the laser scanner system.

Component	Specification
Solenoid coils	150 turns, 2.5 mm Inner Diameter , and 6 mm length
Permanent magnets	Inner Dia. = 0.5 mm, Outer Dia. = 1.0 mm, and 1.0 mm thickness
Photo-detector sensor	PDP90A, 320-1100nm wavelength
Current driver board	Imax = 250mA each channel
Dornier H20 Laser	20 Watts, 2100nm IR beam with 535nm visible light for targeting
Collimating lens	N-BK7 Uncoated Plano-Convex Lens, f=25.4 mm, Dia. = 25.0 mm
Focusing lens	N-BK7 Uncoated Plano-Convex Lens, f = 150.0 mm, Dia. = 25.0 mm

Figure 4.3 The components used by prototype 1 laser scanner system.

voltage, one suitable for designing a closed-loop system. An amplifier circuit board was designed to vary voltage levels and to re-scale voltage into a suitable format for a closed-loop feedback system. In order to generate the desired current for the coils, a current driver board was designed and fabricated, one that would provide the necessary currents to the individual coils, in the N-S-E-W configuration mentioned previously. This system control arrangement allowed the system to scan 2D target surfaces. An Arduino Due microcontroller was used to control the driver board, read the sensor position, and to transfer the processed data to the computer for analysis and for making closed-loop control decisions. A graphical user interface (GUI) was designed to handle the desired value reference signal. Different operating modes and scanning geometries can be programmed into the system via the GUI. A more detailed explanation of the functioning of the system will be provided later in the chapter. Table 4.1 and Figure 4.3 represent the specifications of the tools that have been used to fabricate the prototype of the closed-loop laser scanning system, i.e., prototype 1.

4.4 Bending Moment Equation

The optical fiber bends due to the application of the external torque created by the interaction between the magnetic materials and the magnetic flux generated by the solenoid coils. Consider an optical fiber of length L shown in Figure 4.4. Let τ_m be the applied torque at the free end of the

Figure 4.4 Optical fiber deflection with the external force/ torque. a) Side view, b) Front view.

optical fiber. By using Euler-Bernoulli theory and assuming small deflections of optical fiber, the motion of the optical fiber can be expressed by the differential equation, [Adh12]

$$EI y''''(z, t) + \tau_m y''(z, t) + \rho A \ddot{y}(z, t) = 0 \quad (4.1)$$

where I and E are the moment of inertia and Young's modulus respectively, A is the cross-sectional area of the fiber optic cable, ρ is the radius of curvature, τ_m is the generated and applied external torque, $\dot{y}$ and y' are the partial derivatives of $y(z, t)$ with respect to time (t) and position (z), respectively. For fiber optic beams in which the cross-section are symmetrical and perpendicular to the neutral plane the moment equation of the fiber optic cable can be represented by,

$$EI \frac{d^2 y}{dz^2} = -\tau_m \quad (4.2)$$

4.4.1 Deflection of Fiber Optic by the External Magnetic Field

By placing a magnetic material within a uniform external electromagnetic field a moment is created. This moment produces an induced torque (τ_m) whose motion is directed towards the magnetic field.

$$\tau_m = V_m \vec{M} \times \vec{B} \quad (4.3)$$

where V_m is the total volume of the magnetic material, $\vec{M}$ is the magnetization and $\vec{B}$ is the external magnetic field. By substituting Eq.(4.3) into Eq.(4.2) we have,

$$EI \frac{dy^2}{dz^2} = V_m M B \quad (4.4)$$

By integrating Eq.(4.4),

$$y(z) = \frac{(V_m M B)}{EI} z^2 + C_1 z + C_2 \quad (4.5)$$

Based on Figure 4.4, where torque is distributed over the coated area $(b - a)$, two different boundary conditions apply in order to solve Eq.(4.2): (1) solve for boundary condition 1: $0 \leq z \leq a$, and separately, (2) solve for boundary condition 2: $a \leq z \leq b$, because the active torque changes along the length of the coated area, i.e.:

$$\tau_m = \begin{cases} V_m M B, & \text{if } (0 \leq z \leq a) \\ \frac{V_m M B}{(b-a)}(b-z), & \text{if } (a \leq z \leq b) \end{cases} \quad (4.6)$$

By applying the boundary condition for the fixed end, i.e. $\frac{dy}{dz}|_{z=0}$ and $y(z)|_{z=0}$ for $(0 \leq z \leq a)$ the coefficients C_1 and C_2 become zero, and the deflection is calculated from Eq.(4.7),

$$y(z) = \frac{V_m M B}{EI} z^2 \quad (4.7)$$

By considering Eq.(4.7) and solving for the boundary conditions for $y(z)_{z=a}$ and $\frac{dy(z)}{dz}|_{z=a}$, the deflation for $a \leq z \leq b$ will be:

$$y(z) = \frac{V_m B M}{6EI} \left[\frac{(b-z)^3}{b-a} - 3(b+a)(b-z) + 2b(b+a) - a^2 \right] \quad (4.8)$$

where, the characteristics of the fiber optic cable are as follows, $(b - a) = 10mm$ is the length of the cable covered by the ring of permanent magnets, $250\mu m$ is the cable diameter, and $b = 5cm$ is the length of the cable from its fixed end to its free end.

4.4.2 The Analysis of a Flexible Fiber Optic Cable

In this section the design, analysis, and simulation of an fiber optic cable, i.e., a flexible beam, surrounded by a ring of permanent magnets, i.e., a point mass, and acted on by an external torque is carried out. COMSOL software was again used to conduct a finite element analysis of the fiber optic cable displacement. In order to mathematically model the electromagnetic field and hence derive the desired torque, time domain Maxwell's equations are used:

$$\int_s (\nabla \times b) da = \oint_C b \cdot dl = \mu_0 \left(I_{enc} + \epsilon_0 \frac{d}{dt} \int_s e \cdot \hat{n} da \right) \quad (4.9)$$

where b is the magnetic flux, e is the electric field, I_{enc} is the enclosed current, ϵ_0 is the electric permittivity of free space, and μ_0 is the magnetic permeability of free space. By expressing the Maxwell's equation in frequency domain we have:

$$J = \nabla \times H \quad (4.10)$$

$$B = \nabla \times A \text{ and } B = \mu_0 \mu_r H \quad (4.11)$$

$$E = -j\omega A \text{ and } J = \sigma E + i\omega D \quad (4.12)$$

where, H and E are the magnetic and electrical fields respectively, B magnetic flux density, D displacement current density, σ electrical conductivity, μ_0 magnetic permeability in free space, μ_r

Figure 4.5 3D model simulation of magnetically actuated optical fiber.

Figure 4.6 Induced electromagnetic flux around the optical fiber for a circle pattern.

permeability of the material, ϵ dielectric permittivity and ω is the operating frequency.

The boundary conditions for the flexible beam system being analyzed were derived using a mixture of analytical and numerical methods. Once the boundary conditions were estimated the EM field strength and the torque needed to drive the prototype 1 system were derived. The COMSOL EM field module simulated the field from the four miniature solenoid coils, the ring of permanent magnets attached to the fiber optic cable, and the induced electromagnetic field around the fiber optical cable. A 3D model of the EM fields surrounding the fiber optic cable is shown in Figure 4.5. As can be seen from Figure 4.5, the EM fields generated by the solenoid coils are represented by red lines, the EM field generated by the permanent magnets are represented by blue lines. By controlling the current in the solenoid coils, the necessary torques to control the location of the tip of the fiber optic cable are generated in 2D. By using only two solenoid coils in the N-S-E-W coil configuration in prototype 1 it does control the movement of the fiber optic cable. Results for an induced EM field and the associated generated torque for the circle reference pattern are shown in Figures 4.6-4.7. As can be seen from the results, by controlling the current by two coils at a time, the movement of the fiber optic cable in one Cartesian quadrant can be controlled. Figure 4.7 shows the movement of the fiber optical cable in the x-axis and y-axes respectively.

4.5 The Design and Simulation of the Optic Lenses in Prototype 1

Because all optical materials absorb EM radiation it was important to analyze the performance of the lenses chosen for prototype 1, particularly so for a high-power laser system. Energy absorbed by

Figure 4.7 .Simulation result of magnetically actuated system for optical fiber deflection in circle pattern with frequency 1Hz.

optical materials, such as lenses, can deform the lenses and alter their reflection index. This impacts the laser beam propagation characteristics. Here, the thermal and structural characteristics of the plano-convex optical lenses chosen were studied and analyzed using the optics software package in COMSOL. This section presents the design and thermal analysis of the two plano-convex lenses used on prototype 1.

4.5.1 Plano-Convex Lens Modeling the Focusing

Figure 4.8.(a) shows two plano-convex lenses, the first lens collimates the beam of light propagated by the fiber optic cable and the second lens focuses the beam to a point, i.e., the target. If the output power of the optical fiber is small, then analyzing the thermal effects can be ignored. However, if the output power is large then the optical lenses absorb energy and as a result thermal effects change the characteristics of the lenses, e.g., a focal shift can result and a thermally induced focal shift can change the focus of the beam. In order to analyze and design the lenses COMSOL software was used. Various physics software modules in COMSOL, such as geometric optics, heat transfer in solids, solid mechanics, and moving mesh, were all used in the analysis of the plano-convex lenses chosen for prototype 1. The ray optic software calculated beam trajectories by solving the following differential equations [COM18]:

$$\frac{dq}{dt} = \frac{\partial \omega}{\partial k} \frac{dk}{dt} = -\frac{\partial \omega}{\partial q} \quad (4.13)$$

where, k is wave vector, ω angular frequency, and q ray position. The angular frequency can be

expressed by:

$$\omega = \frac{c|k|}{n} \quad (4.14)$$

where, c is the speed of light and n is the refractive index. By considering Snell's law for the input and output of the lenses we have:

$$n_1 \sin(\theta_1) = n_2 \sin(\theta_2) \quad (4.15)$$

where, θ_1 and θ_2 are the incident and the refraction angles, respectively. Fresnel equations were used to calculate the refraction rays for the high power laser in the test-bed; and coated lenses were chosen to minimize the refraction rays. The propagation of power in a non-absorbing area of the lenses was taken to be constant. In all practical applications high quality optical materials does absorb energy, in this application the energy for each reflection ray ($Q(r)$) was taken to decay exponentially and the following equation was applied:

$$Q(r) = Q(r_0)e^{-2k_0\kappa|r-r_0|} \quad (4.16)$$

where, k_0 is the wave number of ray. The power loss by the reflection rays was solved by implementing the module "Heat transfer in solid" within the the COMSOL software. The power propagation through the lenses was calculated using:

$$\frac{dQ_p}{dt} = \sum_{j=1}^{N_t} \frac{dQ_j}{dt} \delta(r - q_j) \quad (4.17)$$

where, Q_j is the power transmitted by the j th ray, δ Dirac delta function, and N_t total number of rays. The power loss was shown to be uniformly distributed through the lenses as the power of the ray increased. The temperature of the lenses was calculated from:

$$\nabla \cdot (-k \nabla T) = Q_p \quad (4.18)$$

where, k is defined as thermal conductivity of the medium materials. The structural deformation

Figure 4.8 a) The 3D model of the collimating and focusing lenses, b) Ray trajectories for the designed lenses, c) Deposited ray power for the designed lenses, d) Percentage of absorbed power by the designed lenses for two different laser powers.

caused by the increase in temperature was calculated from:

$$\epsilon_{sd} = \alpha(T - T_{ref}) \quad (4.19)$$

where, ϵ_{sd} is the total inelastic strain that occurs in lenses and T_{ref} is the reference temperature. In the COMSOL software the "mechanics interface" module calculates the structural deformation.

4.5.2 Simulation of the Lenses Using COMSOL

In this section the two plano-convex lenses were simulated by the COMSOL software and thermal and the structural deformation were analyzed for a high-power laser system. In this simulation the size of the lenses have been considered the same sizes as required for the application. Also, the lenses were analyzed under 2100nm laser wavelength with maximum 20 Watts power. The Fused Silica material is considered to be the material from which the lenses were constructed. The results are shown in Figures 4.8(b,c, and d). As can be seen from the results, the energy is absorbed by the lenses and converted into heat and this increases the temperature of the lenses themselves. The lenses were tested under two different laser powers: 1Watt and 20 Watts. The results show that the

Figure 4.9 a) The 3D view of the designed current controller board, b) The schematic for the Voltage to current conversion circuit, c) Schematic for the Digital to Analog converter circuit, d) Schematic for the power supply circuit.

designed lenses absorbed energy but still there was no changes in the focal spot of the beam light and as a result these lenses were used for this application.

4.6 Current Controller Board Design

The microcontroller used in this setup is the Arduino Due, it has two Digital to Analog Converters (DAC) so a new current controller board was required, so that it could handle the control current being sent four solenoid coils. Also, because of the maximum current limitations of the Arduino Due microcontroller, a current buffer circuit was designed and fabricated to isolate the solenoid current from the Arduino Due current and therefore to protect the microcontroller. The new prototype 1 controller board can control the currents sent to six individual solenoid coils by generating a maximum current of 250mA for each coil. This maximum current is enough to generate the torques required to control the endoscopic tools fiber optic cable. A detailed explanation of the designed current driver board is as follows:

1. 12-bit DAC. In order to increase both the resolution and the number of the outputs, three DACs were needed. Each DAC unit has two output channels, i.e., six outputs. The circuit schematic is shown in Figure 4.9.(c). Based on the circuit, the SCK signal is the required clock

for DAC at a rate of 4MHz . SDI is the digital value that should be converted into the analog signal and CS1 is the chip select signal that enables the DAC. Both DAC0 and DAC1 generate analog voltages.

2. Current control buffer. Since the DAC can only handle 20mA , an additional buffer part was built to output current. The gain of the circuit is one, and only amplifies the output current; it can handle 1A at the output side. The circuit that handles the output current is shown in Figure 4.9.(b).
3. Power section. This part of the circuit converts 24V to $\pm 9\text{V}$ that is voltage range required to power the Op-Amps chosen. Figure 4.9.(d) shows the power circuit. The PCB of the whole designed circuit is shown in Figure 4.9.(a). The three DACs, and six Op-Amps to handle the current to the coils are shown. Lastly, six 10Ω , 2Watt resistors are used as load in series with the coils.

4.6.1 Shifting and Re-scaling the Output Voltage of the Photo-detector Sensor

The photo-detector sensor used in prototype 1 of the new laser scanner system (endoscopic tool) is a two dimensional, tetra lateral positioning sensor (Appendix D) that provides the position of the beam in the X-axis and the Y-axis. This sensor uses a single resistive layer with a common anode and cathode on each side of the detection area. It measures the position of light by proportionally distributing photo-current when the incident light is applied to its surface. Note that in order to get the best resolution from the sensor there is a SUM voltage that must be approximately $+4\text{V}$, i.e., the best operating voltage for the sensor. This voltage must be checked before use and the light intensity adjusted to ensure a $+4\text{V}$ signal. The output of the sensor will be a voltage between -4V and $+4\text{V}$ for each direction where $+4\text{V}$ is the maximum displacement of light beam at the right, or up side, of the sensor. In order for any new endoscopic tool system to read an output voltage a new re-scaling electronic board design will be necessary. Figure 4.10 shows the new circuit design. This circuit will convert the signal from $\pm 4\text{V}$ to $0-3\text{V}$ which is compatible with the Arduino Due microcontroller. As can be seen, there are two channels that can handle the output of the sensor for both directions. A first order low-pass filter circuit was used to remove the noise from signals and to smooth the output voltage. Note that the values of the low-pass filter should be design (cutoff frequency of the

Figure 4.10 The schematic and the circuit board to re-scale the sensor output to the appropriate signal for the microcontroller.

filter) in such a way that it does not change the phase of the output signal, i.e., it only removes noise.

4.7 Conclusion

In this chapter, the first prototype (prototype 1) of a laser scanner ablation system with closed-loop feedback control was designed, fabricated and analyzed. First, the design of this new laser scanner system was simulated using the COMSOL software, in order to determine the maximum Electro-Magnetic (EM) field required to drive and control the displacement of a laser's fiber optic cable. In addition, the optical lenses used in prototype 1 were modeled and simulated, again by using COMSOL software, to evaluate the light absorption rate of the lenses under a laser's maximum power operating conditions. In the prototype 1 system design an Arduino Due microcontroller was used to control the: (1) output current levels coming from the systems driver board, (2) output signal from the photo-detector sensor, and (3) serial communication. Due to the limitation on the number of DAC channels on the Arduino Due and the maximum output currents, the design of the prototype 1 current driver board had to: (1) control the current of the solenoid coils individually, and (2) generate the maximum output current required to control the displacement of the lasers fiber optic cable in four different directions (N-S-E-W) individually. The driver board was also used to calibrate the photo-detector sensor signal by providing an offset signal. Thereafter, the photo-detector sensor accurately measured the displacement of the laser light beam and to provide the closed-loop control signal that was desired in this new system design.

CHAPTER

5

EXPERIMENTAL RESULTS (PROTOTYPE 1)

5.1 Introduction

In Chapter 4 a closed-loop Electro-Magnetically (EM) actuated laser scanner system was designed and fabricated. In order to accurately control the tip position of the laser's fiber optic cable a controller was designed to compensate for any output tracking error or offsets. The controller design was based on the position of the tip of the fiber optic cable. A photo-detector sensor was used to derive the position of the tip. It was this data that was being fed-back to the controller in order to achieve precise closed-loop positional control of the tip of the endoscopic tool, i.e., prototype 1. This chapter reports on the results obtained from testing prototype 1 when conducting laser ablation tests on biological tissue. The results obtained under open-loop and closed-loop control were analyzed and evaluated. The evaluation of the results led to the design of a new endoscopic tool.

Figure 5.1 Calibration of the open loop system. Left, the position of the beam light in x- and y- axis. Right, The phase diagram of the system in x-y plane.

5.2 Calibration the Open-loop Control System

In the design of the prototype 1 system the laser fiber optic cable was intended to be fixed in-line with the center line of the endoscopic tool's cylinder body, see Figure 5.1. However, due to the permanent magnets being attached to a flexible fiber optic cable, i.e., a flexible cantilever system, there was an offset. This offset was due to the effects of gravity acting on the mass of the permanent magnets on the fiber optic cable, a flexible beam, i.e., a beam that deflects in the manner of a cantilever. From a study of the analysis of statics a cantilever deflects either as a parabola or as a cosine wave. In order to calibrate the fiber optic, i.e., to center it, it was necessary to drive the appropriate solenoid coils in the N-S-E-W configuration by passing an offset current through them. This offset current centered the reflected light from the beam-splitter in the photo-detector. So, to calibrate the system, the current transmitted to the solenoid coils was adjusted until the fiber optic cable was centered. Figure 5.1 shows the open loop response of the system after calibration. As can be seen the system was then able to hold the fiber optic cable at the center of the photo-detector sensor, i.e., the center of the axis. Also, the result shows that the system starts at the origin (0,0) and its positional control was robust enough to return the fiber optic cable to the origin, i.e., the center of the cylinder housing, at the end of the experiment. Although the accuracy of the open loop system was acceptable for tracking step input signals in the x-axis and y-axis, there was an output tracking error. This error occurs because the system was operating under open loop control, so it fails to compensate for any

Figure 5.2 Top, the persistently excited signal applied to open loop system. Bottom, the captured output by photo-detector.

tracking errors.

5.3 System Identification

After calibrating the open loop system, experiments were conducted to find the frequency response of the system. The system order needs to be identified so that the working range of the operating frequency, of the system can be calculated [Moh19a]. An understanding of the speed of response of the system plays an important role in optimizing a system's performance. It means that by finding the maximum operating frequency, the system will be capable of ablating the tissue with minimal thermal damage. In order to get the frequency response, the system was excited at a range of frequencies (up to 20Hz) and with different output geometric patterns. In order to excite all the

Figure 5.3 Box Jenkins system identification model.

frequencies of the system, the input signal was continuously excited. Although the fiber optic was excited at a range of frequencies, the amplitude remained constant. Figure 5.2 shows the range of continuously excited signals used for system identification. In Figure 5.2 the top figure shows the input signals that were applied to the open loop system and the bottom figure shows the output signals gathered by the photo-detector. As can be seen different frequencies with different output patterns such as circles and squares are used to excite all the frequency modes of the system. In order to identify the dynamic behavior of the system, a Box-Jenkin identification model was used. Figure 5.3 shows the general model of Box-Jenkins identification approach. $u(k)$ and $y(k)$ are the input and output sequence of data, and $e(k)$ is white noise sequence data. The polynomial functions $C(z)$, $D(z)$, $B(z)$, and $F(z)$ are defined as:

$$\begin{aligned}
 F(z) &= 1 + f_1 z^{-1} + f_2 z^{-2} + \dots + f_{n_f} z^{-n_f} \\
 B(z) &= 1 + b_1 z^{-1} + b_2 z^{-2} + \dots + b_{n_b} z^{-n_b} \\
 C(z) &= 1 + c_1 z^{-1} + c_2 z^{-2} + \dots + c_{n_c} z^{-n_c} \\
 D(z) &= 1 + d_1 z^{-1} + d_2 z^{-2} + \dots + d_{n_d} z^{-n_d}
 \end{aligned} \tag{5.1}$$

and:

$$\begin{aligned}
 x(k) &= \frac{B(z)}{F(z)} u(k), \text{ and } w(k) = \frac{C(z)}{D(z)} e(k) \\
 y(k) &= x(k) + w(k)
 \end{aligned} \tag{5.2}$$

By defining the vector parameters:

$$\begin{aligned}
 \theta &= [\theta_i \ \theta_n] \\
 \theta_i &= [f_1 \ f_2 \ \dots \ f_{n_f} \ b_1 \ b_2 \ \dots \ b_{n_b}] \\
 \theta_n &= [d_1 \ d_2 \ \dots \ d_{n_d} \ c_1 \ c_2 \ \dots \ c_{n_c}]
 \end{aligned} \tag{5.3}$$

$$\begin{aligned}
\phi(k) &= [\phi_i \ \phi_n]^T \\
\phi_i(k) &= [-x(k-1) \cdots x(k-nf) \ u(k-1) \cdots u(k-nb)] \\
\phi_n(k) &= [-w(k-1) \cdots w(k-nd) \ e(k-1) \cdots e(k-nc)]
\end{aligned} \tag{5.4}$$

By using Eq.(5.2), Eq.(5.3) and Eq.(5.4) we have:

$$\begin{aligned}
x(k) &= \phi_i^T(k) \theta_i \\
w(k) &= \phi_n^T(k) \theta_n + e(k) \\
y(k) &= \phi_i^T(k) \theta_i + \phi_n^T(k) \theta_n + e(k) = \phi(k) \theta + e(k)
\end{aligned} \tag{5.5}$$

The goal is to find the unknown vectors $\phi(k)$ and $\theta(k)$. The dynamic model of the system and noise can be derived from these vectors. In order to estimate these unknown parameters, based on the measured data, a cost function must be defined and optimized. The cost function was derived using a least squares error approach, by comparing the error between the output of the physical system and the estimated output from the model,

$$J(\theta) = \sum_{i=1}^l [y(i) - \hat{\phi}(i)\theta]^2 \tag{5.6}$$

Figure 5.4 Frequency response of the identified model.

Figure 5.5 The estimate and actual outputs of the system.

By solving this optimization problem, using numerical methods, the following recursive formula is derived to update the parameters,

$$\hat{\theta}(k) = \hat{\theta}(k-1) + P(k)\hat{\phi}[y(k) - \hat{\phi}^T(k)\hat{\theta}(k-1)] \quad (5.7)$$

$$P(k) = P(k-1) - \frac{P(k-1)\hat{\phi}(k)\hat{\phi}^T(k)P(k-1)}{1 + \hat{\phi}^T(k)P(k-1)\hat{\phi}(k)} \quad (5.8)$$

$$\hat{x}(k) = \hat{\phi}_i(k)\hat{\theta}_i(k)$$

$$\hat{w}(k) = y(k) - \hat{x}(k) \quad (5.9)$$

$$\hat{e}(k) = \hat{w}(k) - \hat{\phi}_n^T(k)\theta_n(k)$$

MATLAB software was used to solve Eqs.(5.7)-(5.8). The order of the numerator and denominator for the Box-Jenkins model are defined by the user. The system model, identified by the Box-Jenkins software, is a second order system. The system model is derived from the data displayed in Figure 5.4:

$$G(s) = \frac{Y(s)}{U(s)} = \frac{13.86s + 9.9e4}{s^2 + 27.7s + 9.889e4} \quad (5.10)$$

Based on the result, the natural frequency of the system is approximately 46.9447Hz. As the sys-

tem excitation frequency gets to the natural frequency the magnitude of the output signal increases. In the actual system, when the excitation frequency is really close to the natural frequency of the system; the system becomes unstable. Here, the frequency response of the system is derived to be second order, so the magnitude around 50Hz does not exponentially increase. The estimated output and the actual output of the system are shown in Figure 5.5. As can be seen from the results the estimated model approximately follows the actual system's behavior.

5.4 Controller Design

After estimating the dynamic model of the laser scanner system, it was then necessary to design a controller for the closed-loop system. The controller must compensate for the tracking error and alter the transient and steady state responses of the system based on the desired objectives. The output laser beam position was measured by the photo-detector sensor and compared with the reference position. If there is any error between the reference signal and the actual position, the controller will calculate the required command and send an error signal to the driver board to compensate for the position error. The controller used in the laser scanner system is a combination of feed-forward controller, to smooth the reference signal, and a Proportional-Integral (PI) controller, to compensate for the tracking error. Figure 5.6 shows the general view of the closed-loop laser scanner system. As can be seen two different controllers were used to handle the movement of the optical fiber and as a result control the position of laser beam. The controller was designed based on the estimated transfer function, Eq.5.10. The PI controller can be designed using different methods in the frequency and time domains. Here the MATLAB sisotool platform was used to design the controller. The objective was to achieve the fastest rise time and settling time with the lowest

Figure 5.6 Block diagram of the closed-loop magnetically actuated laser scanner system.

Figure 5.7 Designed graphical user interface to handle the laser scanner system.

overshoot (less than 10%). The designed controller parameters are as $K_p = 16$ and $K_I = 0.08$.

5.5 Testing and Evaluating the System

The system, as designed, is tested and evaluated under different scenarios. A Holmium H20 Laser was used for ablation experiments [Moh19b]. The maximum power of the laser is 20 Watts at a maximum 12Hz operation frequency. The H20 laser is a pulsed laser, which is commonly used for fragmenting kidney stones. The laser ablation prototype system was tested on wood and chicken tissue using different geometric patterns. A graphical user interface (GUI) was designed to help the user send the reference patterns (commands) to the system. Figure 5.7 shows the GUI designed for the laser scanner system. As can be seen patterns to be tested were placed in the GUI, making it easier for the user to operate. There is also a Joystick mode for the user to control ablation, i.e., open-loop, manual control. Also, the GUI is capable of hosting MRI images and segmenting a tumor (see Figure 5.7). After segmenting the image and extracting the coordinates of the tumor, an appropriate user command to the controller activates a raster scan ablation, or a border line ablation, whichever is appropriate [Moh18]. In the first evaluating trial the Holmium H20 laser was initialized to have a visible light intensity that was suitable for the photo-detector sensor. All the tests were undertaken at the Wake Forest Baptist Medical Center (WFBMC), Winston-Salem, NC. The WFBMC provided the Dornier H20 laser for evaluating the laser scanner system and technical

Figure 5.8 Dornier H20 operation interface.

support. Before adjusting the intensity the user must check that the fiber cable is detected by the laser, or not. There is a warning message by the interface that shows if the fiber cable is detected by the laser system or not. In order to tune the intensity, the SUM output signal of the photo-detector should be measured. The required output voltage for this signal should be set to $+4V$. The user should adjust the intensity of the visible light until the SUM voltage reaches $+4V$. The resolution will also change based on an incorrect voltage being adjusted. A voltage of more than $+4V$ will saturate the photo-detector sensor and impact the signals being recorded.

After adjusting the intensity of the visible light, the user must then tune the IR laser beam to its required operating power. Both frequency and energy options will change the output power of the laser. Figure 5.8 shows the screen of the Holmium 20 laser with different setting parameters. As can be seen the intensity of visible light can be tuned by pilot option. The output power can be increased by adjusting the frequency and energy, or both.

5.5.1 Evaluating the Closed-loop Laser Scanner System

The closed-loop laser scanner system was tested to create different geometric shapes. The IR laser in this case was off and only visible light was used to evaluate the closed-loop system. The reflected beam light was measured by photo-detector and feedback to the controller to reduce the tracking error. The controller also is able to transfer the actual position of the optic laser to the GUI through serial communication. Different scenarios were considered to evaluate the system under different amplitudes and operation frequencies. Figures 5.9-5.11 show the output of the closed-loop laser scanner for different geometric shapes. The Mean Square Error (MSE) of the output signal for the designed system is about $194\mu m$ however, the thickness of the beam has not been considered. As it can be seen from the results the system is capable of operating accurately at high-speed and small

Figure 5.9 Output of the system for circle pattern.

Figure 5.10 Output of the system for square pattern.

amplitudes. The Root Mean Square Error (MSE) and standard deviation (SD) of the output error have been summarized in Table 5.1.

5.5.1.1 Effect of Increasing the Frequency

In this case the system was tested under circle and square patterns with operation frequencies 5Hz and 8Hz. Based on the estimated model the designed system is capable to create a pattern with maximum 10Hz frequency. The results have been shown in Figures 5.12-5.15. The system is capable of tracking the reference patterns with good performance.

Figure 5.11 Output of the system for infinity pattern.

Figure 5.12 Output of the system for a circle pattern with 5Hz frequency.

Figure 5.13 Output of the system for a circle pattern with 8Hz frequency.

Figure 5.14 Output of the system for a circle pattern with 5Hz frequency.

Figure 5.15 Output of the system for a circle pattern with 8Hz frequency.

5.5.1.2 Effect of Different Amplitudes

In this scenario the system was tested for a circle pattern with different amplitudes. The result is shown in Figure 5.16. As can be seen the system is capable of tracking the reference pattern with good performance.

5.5.2 Evaluating the System on Wood Material

After adjusting all settings, the laser scanner system was used to create geometrical patterns on wood. Figure 5.17 shows the output of the laser scanner system for different geometrical patterns. As can be seen the laser scanner system was capable of burning patterns into the wood with a variety of

Figure 5.16 Output of Laser scanner system for circle patter with different diameter.

Table 5.1 Root Mean Square Error (RMSE) and Standard Deviation (SD) of the output error signals (Prototype 1).

Pattern Type	RMSE	SD
Circle Pattern (f = 1Hz)	194 μm	192 μm
Circle Pattern (f = 5Hz)	139 μm	132 μm
Circle Pattern (f = 8Hz)	93.5 μm	89 μm
Circle Pattern (f = 1Hz)		
Amplitude: 2.0mm	193 μm	192 μm
Amplitude: 1.5mm	121 μm	109 μm
Amplitude: 0.5mm	67.2 μm	55 μm
Square (f = 1Hz)	178 μm	112 μm
Infinity (f = 1Hz)	183 μm	184 μm

shapes and different amplitudes. Also the speed of operation can be under user control. The power of the laser was adjusted to carbonize the required shape only. Due to the laser operating restrictions, laser power was not tested at its maximum output power. Note that since the Dornier H20 laser is a pulse laser only, the frequency operation of the laser was the key factor during the burning the wood and creating the patterns.

5.5.3 Evaluating the System on Chicken Tissue

In the second set of experiments the laser scanner system was tested and evaluated on chicken tissue to evaluate the performance of the laser scanner system for biological tissue ablation. Similar

Figure 5.17 The output of the laser scanner system tested on wood material.

Figure 5.18 Evaluating the laser scanner system on chicken biological tissue.

to the experiments with wood, a variety of geometric shapes were generated on the chicken tissue at different levels of power. The results are shown in Figure 5.18. As can be seen the system was capable of creating the patterns on the chicken tissue. Since the Dornier H20 laser was built to be used for fragmenting kidney stones, the system was not able to ablate the tissue patterns precisely, even with increased laser output power being applied. Increasing the output power here gave a heat distribution in the neighborhood of the pattern created in the tissue until it eventually gave a heat concentration in the center of the created pattern. This effect would have ultimately resulted in a carbonized center of the pattern.

5.6 Conclusion

In Chapter 5, the prototype 1 system designed in Chapter 4 was tested and its performance analyzed. The experiments carried out at the Wake Forest Baptist Medical Center. The experiments used a Dornier H20, pulsed laser system. The fiber optical cable for this system carried both: (1) an infrared

(IR) laser light, with a 2100nm wavelength, and (2) a visible green light, with a 535nm wavelength. The visible green light was used first, to aid position targeting, then the IR light was used for laser ablation experiments. First, this new system required calibration. This was achieved by determining the external torque necessary to hold the light beam steady at a desired target location, i.e., by compensating for the gravitational effects acting on the fiber optic cable. Second, in order to design an appropriate system controller, the order of the system had to be determined. The system order was determined by applying the Box-Jenkins identification model, which derived the system order as being second-order. In physical experimentation a range of excitation signals was applied to the prototype 1 system, i.e., a range of excitation signals with different amplitude and frequencies. This approach determined the working range of the system and the modes of vibration of the system. The experimental results validated the results obtained from the Box-Jenkins identified model. Based on these results a PI controller was designed and applied to the prototype 1 system. The output from the feedback control loop was then added to the output from the feed-forward controller in order to generate the appropriate current signals that are sent to the individual coils (N-S-E-W) that control the displacement of the endoscopic tools fiber optic cable. A beam splitter was introduced both to transmit the IR beam to the target and to reflect the visible light for feedback loop light intensity measurement. The reflected beam was measured by a photo-detector sensor and fed back to the controller to reduce the tracking error. The photo-detector sensor measures the actual position of the laser light beam during ablation testing and feeds this measured signal back to the system in order to improve the tracking error. The fabricated laser ablation system was evaluated on two different materials, i.e., wood and biological tissue. The physical experimental results for ablation patterns, generated for a range of frequencies and amplitudes, showed that the prototype 1 laser scanner system gave consistent, robust, and accurate ablation patterns, with both wood and biological tissue. However, an ultra-fast pulsed laser system is required to ablate the biological tissue with minimal thermal expansion. Also, the frequency response experiments conducted with prototype 1 showed that the system is able to ablate biological tissue robustly, accurately and consistently.

CHAPTER

6

THE DESIGN, FABRICATION AND TESTING OF THE LASER SCANNER BIOLOGICAL TISSUE ABLATION SYSTEM – PROTOTYPE 2

6.1 Introduction

In chapters 4 and 5 the first prototype laser scanner system was designed, implemented and tested. Experiments with prototype 1 for ablating biological tissue with different power levels and geometric patterns, using a pulsed laser, showed that there were problems with the type of laser which this causes heat distribution in the tissue. Furthermore, the designed prototype 1 system should be scaled-down and the proposed control approached must be evaluated with the scaled-down system. This chapter focuses on design of the new endoscopic tool, i.e., prototype 2. The design and

Table 6.1 The specification of the components used for the scaled-down laser scanner system.

Component	Specification
Solenoid coils	300 turns, 1.0 mm inner Dia., 1.5mm outer Dia., and 10mm length
Permanent magnets	Inner Dia. = 0.5 mm, Outer Dia. = 1.0mm, and 1.0mm thickness
Photo-detector sensor	PDP90A, 100-1100nm wavelength
Current driver board	Imax = 250mA each channel
Diode Laser	Max: 34 Watts, 1470 nm IR beam with 635nm pilot visible light
Collimating lens	N-BK7 Uncoated Plano-Convex Lens, Dia. 6.0 mm, f = 10 mm
Focusing lens	N-BK7 Uncoated Plano-Convex Lens, Dia. 10.0 mm, f = 100.0 mm

scale of prototype 2 will address the problems highlighted in the experiments with prototype 1. In addition, the experiments with prototype 2 will be carried out with a continuous wave laser. This new laser scanner system will give more accurate and precise biological tissue ablation, because of the endoscopic tool design changes and because of its closed-loop feedback. Having closed-loop control embedded into the new laser scanning endoscopic tool system takes medical procedures a step closer to achieving the automation ablation of biological tissue. The manufactured system is tested and evaluated by a diode laser with 1470nm IR beam and 635nm red beam as a pilot beam.

6.2 Prototype 2 Design

As mentioned before, several constraints should be considered to design a magnetically actuated laser scanner system. In order to scale down the system, the electromagnetic coils that are used to generate the required EM fields, should be designed properly to generate an adequate amount

Figure 6.1 The designed components used for the scaled-down laser scanner system.

Figure 6.2 General view of the prototype system 2. Photo-detector sensor (1), Diode Laser system (2), Designed Laser scanner system (3), Laser system Driver Board (4), Laser scanner Driver Board.

of torque to be able to control the movement of the fiber optic. In this prototype, compare to the previous system, the new designed coils have 300 turns, length of 10mm and inner diameter of 1mm . The optical lenses were considered to have the diameter of 6mm and 10mm and are capable to handle the 34 Watts power in 1470nm wavelength. Figure 6.1 shows some of the parts that are manufactured for this test-bed. The specification of all the parts that are used for the new prototype system is summarized in Table. 6.1 and Appendix B. The general view of the designed system has been shown in Figure 6.2. The optical fiber is fixed at one side of the cylinder and five permanent magnets with the same specification as the previous prototype are used to be placed around the fiber optic. As can be seen from Figure 6.2, the beam splitter is attached at the end of the main tube case in order to split the visible light that is used for the feedback loop to be added to the feed forward signal and improves the performance of the system output. The outputs of the photo-detector sensor are corresponding to the difference position of the beam light in x – and y – directions respectively. An analog to digital converter (ADC) shield is used to read the output signals of the sensor and apply them to the control loop. The currents of the coils are controlled using the designed current driver board which is explained in detail in Chapter 4.

Figure 6.3 a) System response for the calibration procedure. b) The phase plot of the system behavior.

6.3 Calibration and System Identification

In this section the same procedure is considered for the new designed system and the system is calibrated and the model of the system is identified by applying proper input signals. The estimated transfer function of the system is derived by applying an appropriate persistently exciting input signals. The behavior of the system after the calibration process has been shown in Figures 6.3(a and b). As it can be seen the system returns to the start point since the require torque power is applied to make sure the system is stopped at the origin (0,0) position without any offset. In order to estimate a precise model of the system, a persistently exciting signal with several frequencies and amplitudes is applied to the designed system and the output position of the system is measured using the photo-detector sensor. The input signal and the output response of the system are shown in Figure 6.4. Similar to the first prototype system, the Box-Jenkins identification approach is used to identify the transfer function of the system. This model considers an objective function based on the estimation error and tries to identify the system's parameters in a way that minimizes the objective function. A numerical and recursive technique is used to identify the parameters. More detail of the system identification approach can be found in Chapter 5. The identified second order model of the system is as,

Figure 6.4 A persistently exciting input/output signals used to identify the system parameters.

$$G(s) = \frac{-176.6s + 2.532e^4}{s^2 + 7.147s + 3.107e^4} \quad (6.1)$$

The frequency response of the identified system is shown in Figure 6.5. The resonance frequency of the system is about 30Hz, however the actual system is capable of performing the maximum 8Hz for the input reference signal.

Figure 6.5 The Bode diagram of the identified model for the second prototype system.

Figure 6.6 The output position of the laser scanner system for 1Hz circle pattern.

6.4 Test and Evaluation of the Designed System

In this section the designed scaled-down system is tested and evaluated under several scenarios. A diode Laser (Appendix C) with the operating wavelength at $1470nm$ is used for evaluating the constructed system for all experiments. The maximum power of the laser is 34 Watts and the laser system is capable of working in three different modes such as continuous wave (CW) mode, pulsed wave mode, and number of shots mode. The experiments have been done in CW operation mode. A pilot red beam light with $635nm$ wavelength is used to be applied to the closed loop system. The results of the system for different frequencies and amplitudes have been shown in Figures 6.6-6.12. As it can be seen the scaled-down system is capable of operating at different frequencies to follow the reference pattern with high performance. Figure 6.13 shows the output response of the system for circle pattern with different amplitudes. The MSE and SD of the system results have been summarized in Table 6.2. The developed system is also evaluated on the biological tissue (chicken tissue) and the results have been shown in Figure 6.14. As it can be seen the heat propagation through the tissue is the main reason that does not allow to create a precise pattern similar to the generated one by the laser scanner on the biological tissue. In order to verify the results, a COMSOL Multiphysics computational package is used to simulate and study the heat propagation through the biological tissue. More extensive results are shown in Appendix A.

The generated heat by the laser beam is radially transferred through the biological tissue based on its thermal conductivity. By continuous beam incident, the temperature of the biological tissue will reach to the critical temperature, $60^{\circ}C$, and as a result coagulation of biological tissue will be

Figure 6.7 The output position of the laser scanner system for 1Hz Infinity pattern.

Figure 6.8 The output position of the laser scanner system for 1Hz Square pattern.

Figure 6.9 The output position of the laser scanner system for 3Hz circle pattern.

happened. More energy will increase the temperature until the boiling temperature 100°C and the water molecules will be evaporated and the biological tissue will start to be shrank and carbonized [Ans13]. A finite element analysis can be applied to model the time-dependent heat propagation through the biological tissue based on the amount of a continuous wave and pulsed laser energy.

Figure 6.10 The output position of the laser scanner system for 5Hz circle pattern.

Figure 6.11 The output position of the laser scanner system for 3Hz square pattern.

Figure 6.12 The output position of the laser scanner system for 5Hz square pattern.

The thermal propagation of the biological tissue can be described by:

$$\rho C_p \frac{\partial T(r, z, t)}{\partial t} - \Delta(k \Delta T(r, z, t)) = Q_{ex} \quad (6.2)$$

Figure 6.13 The output position of the laser scanner system for 1Hz circle pattern with 0.5mm, 1mm, and 2mm amplitude.

Table 6.2 Root Mean Square Error (RMSE) and Standard Deviation (SD) of the output error signals (Prototype 2).

Pattern Type	RMSE	SD
Circle Pattern (f = 1Hz)	40.0 μm	33.9 μm
Circle Pattern (f = 3Hz)	37.8 μm	31.5 μm
Circle Pattern (f = 5Hz)	38.7 μm	38.0 μm
Circle Pattern (f = 1Hz)		
Amplitude: 2.0mm	40.0 μm	33.9 μm
Amplitude: 1.5mm	25.6 μm	21.0 μm
Amplitude: 0.5mm	20.2 μm	15.4 μm
Square (f = 1Hz)	124.7 μm	72.6 μm
Infinity (f = 1Hz)	107.5 μm	101.7 μm

where ρ is the density, k is the thermal conductivity of the biological tissue, C_p is the heat capacity for biological tissue, z is the coordinate along incident beam, and Δ is the gradient operator. Q_{ex} represents the total heat generated by the incident laser beam in tissue. T and t express the temperature and time respectively. The laser energy is modeled by Beer-Lambert Law as:

$$I(r, z) = I_0 e^{-(\mu_a + \mu_s)z}$$

$$Q_{beam} = -\frac{\partial I(r, z)}{\partial z} = \mu_a I(r, z) \quad (6.3)$$

where I_0 is the intensity of laser beam at the surface of the biological tissue, μ_a and μ_s are the

Figure 6.14 a) Experimental result of the laser scanner system for circles and square patterns with 1 Watt intensity power. b) Experimental results of the laser scanner system for different patterns on chicken biological tissue. c) Experimental result of the laser scanner system for circles, square, infinity and eight patterns.

absorption and scattering coefficients of the biological tissue. The thermal radiation and heat loss from the tissue surface which has contact with the environment is modelled by the diffusion equation:

$$-n.(-k\Delta T(r, z, t)) = \epsilon\sigma(T_{amb}^4 - T^4) \quad (6.4)$$

where ϵ is the surface emissivity and $\sigma = 5.67 \times 10^{-8} \text{ W} \cdot \text{m}^{-2} \cdot \text{K}^{-4}$ is the Stefan-Boltzmann constant. The simulated results from COMSOL model for CW laser with 1470 nm wavelength and 15 Watts power have been shown in Figures 6.15 (a,b,c,d). COMSOL Multiphysics package performs Eqs.6.2-6.4 to solve the temperature distribution in biological tissue. A muscle biological tissue is modeled by COMSOL software with properties such as $k = 0.49 \text{ W}/(\text{m} \cdot \text{K})$ as thermal conductivity, $\rho = 1090 \text{ kg}/\text{m}^3$ and $c_p = 3421 \text{ J}/(\text{kg} \cdot \text{K})$ as density and heat capacity respectively. As can be seen from Figure 6.15, although the beam diameter is $250 \mu\text{m}$, the heat propagates through the biological tissue with a thickness more than 1 mm with high temperature which either coagulates or carbonizes the biological tissue and destroys the desired pattern. In order to find a solution that helps to prevent

Figure 6.15 Heat propagation Simulation result for biological tissue (Muscle) for different scanning patterns: a) Circle Pattern, b) Square Pattern, c) Infinity Pattern, d) Eight Pattern.

the high temperature distribution through the biological tissue, a high peak power pulsed laser with nanosecond pulse width is modeled and simulated using COMSOL Multiphysics package and the simulations are compared with the CW laser simulation outcomes. Figure 6.16(a,b,c,d) shows the simulation results for this scenario. Figure 6.16.(b) and Figure 6.16.(d) show the temperature distribution on the surface of the biological tissue for CW laser and pulsed laser respectively. The results show that the pulsed laser with high peak power has more focused heat profile than the CW laser system. As a result with a ultra-fast pulsed laser, significant improvement in pattern ablation with finer ablation path and less damage to the biological tissue can be achieved.

Figure 6.16 Heat propagation through the biological tissue (Muscle) for a Continuous laser (CW) and Pulsed laser System: a) 3D model of CW laser system, b) The temperature of Biological tissue vs. surface width for CW laser system, c) 3D model of Pulsed laser system, b) The temperature of Biological tissue vs. surface width for Pulsed Laser system.

6.5 Conclusion

Based on the results obtained with prototype 1, Chapter 5, a new scaled-down design of a magnetically actuated laser scanner system was developed and tested. This new system, termed prototype 2, not only included a new endoscopic tool design, it was also constructed using a continuous wave laser. The experiments in Chapter 4, with the Dornier H20, pulsed laser system did not give an optimized performance. The design of laser scanner system of prototype 2 incorporated a diode

laser, with 34 Watts power, 1470nm IR beam wavelength and 635nm red light beam specifically purchased for this new system. For laser scanner system light beam displacement the four miniature electromagnetic (EM) coils were again controlled individually from currents generated by the driver board. The torque needed to displace the fiber optic cable is generated from the interaction between the EM coils hosted in the endoscopic tool and the permanent magnets hosted on the flexible fiber optic cable. Experimental results show that the design of the scaled-down laser scanner system was capable of tracking a variety of geometric reference patterns, with high accuracy and speed. The developed system was tested with a biological tissue. The experimental results showed that the laser scanner system generated heat that quickly propagated through the biological tissue, such that this heat coagulated and carbonized the biological tissue close to the site of the geometric pattern being generated. In order to analyze this heat transmission effect more closely a COMSOL model was constructed for a laser scanner system with a continuous wave (CW) laser ablating biological tissue. The simulation results show that where thermal conductivity of the biological tissue is in fact when the heat propagates through the biological tissue rapidly, it does coagulate and carbonize the site surrounding the geometric pattern inscribed by the laser light. The COMSOL model shows that in order to reduce the heat propagation effect on biological tissue, a nanosecond pulsed laser system should be used in a laser scanner system. Trials with the COMSOL model a nanosecond pulsed laser system were compared with the CW laser model used for prototype 2. COMSOL simulation results also show that by applying an ultra-fast laser scanner system, in association with a fiber optic cable of 250 μ m diameter; precise biological tissue ablations can be achieved with minimal thermal damage occurring at the tissue sites surrounding the ablation site.

CHAPTER

7

CONCLUSION

The origins of the research in this dissertation came from two separate research fields: (1) the design, modeling, and physical testing of a new wireless inductive proximity sensor system for detecting hip joint implant loosening, and (2) by studying the European Union μ Ralp research project into developing new laser-based endoscopic tools for tissue ablation surgery; which was an extension of research into biological tissue ablation by Giallo and Grant at NC State University.

First, it became apparent that the research into the new wireless inductive proximity sensor system for detecting hip joint implant loosening could be adapted to the laser ablation task. By varying the current being applied individually to four solenoid coils and attaching permanent magnets to a laser fiber optic cable, the displacement of the fiber optic cable could be accurately controlled. Second, if a photo-detector sensor was added to the system, to measure the reflected visible beam, a closed-loop feedback control system would achieve. A search of the available literature showed that no other such laser research test-bed in biomedical tissue ablations, commercial or otherwise, appeared to possess closed-loop feedback control of laser light beam. The first prototype (prototype 1) of a laser scanner ablation system with closed-loop feedback control was designed, fabricated

and analyzed. First, the design of this new laser scanner system was simulated using the COMSOL software, in order to determine the maximum electromagnetic (EM) field required to drive and control the displacement of a laser's fiber optic cable. In addition, the optical lenses used in prototype 1 were modeled and simulated, again by using COMSOL software, to evaluate the light absorption rate of the lenses under a laser's maximum power operating conditions. In the prototype 1 system design, an Arduino Due microcontroller was used to control the: (1) output current levels coming from the systems driver board, (2) output signal from the photo-detector sensor, and (3) serial communication. Due to the limitation on the number of DAC channels on the Arduino Due and its maximum output current, the design of the prototype 1 current driver board had to: (1) control the current of the solenoid coils individually, and (2) generate the maximum output current required to control the displacement of the lasers fiber optic cable in four different directions (N-S-E-W) individually. The driver board was also used to calibrate the photo-detector sensor signal by providing an offset signal. Thereafter, the photo-detector sensor accurately measured the displacement of the laser light beam and to provide the closed-loop control signal that was desired in this new system design. Once constructed the functional of the prototype 1 scanning system was tested and its performance analyzed. The experiments were carried out at the Wake Forest Baptist Medical Center. The experiments used a Dornier H20, pulsed laser system. The fiber optical cable for this system carried both: (1) an infrared (IR) laser light, with a $2100nm$ wavelength, and (2) a visible green light, with a $535nm$ wavelength. The visible green light was used first, to aid position targeting, then the IR light was used for laser ablation experiments. First, this new system required calibration. This was achieved by determining the external torque necessary to hold the light beam steady at a desired target location, i.e., by compensating for the gravitational effects acting on the fiber optic cable. Second, in order to design an appropriate system controller, the order of the system had to be determined. The system order was determined by applying the Box-Jenkins identification model, which derived the system order as being second-order. In physical experimentation, a range of excitation signals was applied to the prototype 1 system, i.e., a range of excitation signals with different amplitude and frequencies. This approach determined the working range of the system and the modes of vibration of the system. The experimental results validated the results obtained from the Box-Jenkins identified model. Based on these results, a PI controller was designed and applied to the prototype 1 system. The output from the feedback control loop was then added to the

output from the feed-forward controller in order to generate the appropriate current signals that are sent to the individual coils (N-S-E-W) that control the displacement of the endoscopic tools fiber optic cable. A beam splitter was introduced both to transmit the IR beam to the target and to reflect the visible light for feedback loop light intensity measurement. The reflected beam was measured by a photo-detector sensor and fed back to the controller to reduce the tracking error. The photo-detector sensor measures the actual position of the laser light beam during ablation testing and feeds this measured signal back to the system in order to improve the tracking error. The fabricated laser ablation system was evaluated on two different materials, i.e., wood and biological tissue. The physical experimental results for ablation patterns, generated for range of frequencies and amplitudes, showed that the prototype 1 laser scanner system gave consistent, robust, and accurate ablation patterns, with both wood and biological tissue. Also, the frequency response experiments conducted with prototype 1 showed that the system ablated biological tissue robustly, accurately and consistently.

Based on the results obtained with prototype 1, a second scaled-down design of a magnetically actuated laser scanner system was designed, developed and tested. This new system, termed prototype 2, not only included a new endoscopic tool design, it was also constructed using a continuous wave laser. The experiments in Chapter 4, with the Dornier H20, pulsed laser system did not give an optimized performance. The design of laser scanner system of prototype 2 incorporated a diode laser, with 34 Watts power, 1470nm IR beam wavelength and 635nm red light beam specifically purchased for this new system. For laser scanner system light beam displacement the four miniature electromagnetic (EM) coils were again controlled individually from currents generated by the driver board. The torque needed to displace the fiber optic cable is generated from the interaction between the EM coils hosted in the endoscopic tool and the permanent magnets hosted on the flexible fiber optic cable. Experimental results show that the design of the scaled-down laser scanner system was capable of tracking a variety of geometric reference patterns, with high accuracy and speed. The developed system was tested with biological tissue. The experimental results showed that the laser scanner system generated heat that quickly propagated through the biological tissue, such that this heat coagulated and carbonized the biological tissue close to the site of the geometric pattern being generated. In order to analyze this heat transmission effect more closely, a COMSOL model was constructed for a laser scanner system with a continuous wave (CW) laser ablating biological tissue.

The simulation results show where and how thermal conductivity of the biological tissue issues occur; in fact this phenomenon appears to occur when heat propagates through the biological tissue too rapidly, i.e., the laser light coagulates and carbonizes the site surrounding the geometric pattern inscribed by the laser light.

The major conclusions drawn from this research are as follows. The endoscopic tool design in the prototype 1 test-bed showed that it was capable of generating relatively accurate geometric patterns, under feedback control using a photo-detector sensor. After calibration, this was achieved by controlling the current in the four-solenoid coils individually, thereby generating the desired control torque by altering the electromagnetic field. Tests on biological tissue with prototype 1 showed that even though laser beam lines drawn on the biological tissue were relatively accurate, the heat generated by the pulsed laser beam being used propagated throughout the surrounding tissue. If prototype 1 were being used on a precise ablation procedure, e.g., a laryngeal tumor removal, the heat propagation into the healthy tissue would most certainly increase the ablation margins. That is, healthy tissue would be destroyed. However, the prototype 1 system with the pulsed laser could possibly be used in tattoo removal or for removing burned skin in dermatological procedures, i.e., where healthy tissue margins are less of an issue. The prototype 2 endoscopic tool design used the same solenoid coil control arrangement, but it was reduced in size and it was modular in construction. It was considered that a modular design would allow new solenoid coil arrangements and control strategies to be tested. The major difference in prototype 2 was the decision to use a continuous wave (CW) laser for the biological tissue ablation experiments. Would endoscopic tool laser beam speed control along with a CW laser eliminate the heat propagation seen in experiments with prototype 1? It was found that although the laser beam position and speed control prototype 2 gave geometric patterns that were more accurate precise within the systems working range, heat propagation still existed; but to a lesser extent. Finding now evidence in the literature addressing endoscopic tool/laser pairing studies on biological tissue a decision was made to develop a COMSOL model of the prototype 2 test-bed. COMSOL simulations gave similar results to those found in physical experiments with biological tissue, i.e., heat propagation in tissue. Further, simulations with the COMSOL model indicated that precise ablation patterns could be generate with almost no heat propagation. However, to achieve this goal an endoscopic tool would have to be used in association with a high-frequency pulsed laser system, e.g., system generates ultrashort

pulses with high power level and less than nanosecond durations.

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APPENDICES

APPENDIX

A

ADDITIONAL EXPERIMENTAL RESULTS

This section presents additional experimental results for the designed prototype 2 test-bed system. Additional experimental result for the chicken tissue and wood material have been shown in Figures A.1-A.2. Figure A.3 shows the data captured by the photo-detector sensor for different output patterns that is used for the feedback loop to improve the tracking error. The output of the diode laser for different output powers and repetition has been shown in Figure A.4. The results show that how the heat is propagating through the chicken biological tissue by increasing the laser power and the repetition rate. Also, it indicates that how much power is required to ablate the chicken tissue.

Figure A.1 Additional Experimental Results on wood material.

Figure A.2 Additional Experimental Results on Chicken tissue.

Figure A.3 The output signal of the photodetector sensor. a) Circle pattern, b) Square pattern, c) Infinity pattern, d) Eight pattern.

Figure A.4 The output of diode laser for different output powers and repetition path. From left to right: 1, 3, 5, and 10 repetition times) a) 3 Watts output power b) 5 Watts output power, c) 10 Watts output power, d) 12 Watts output power.

APPENDIX

B

MECHANICAL DESIGN OF LENS HOLDERS, BEAM SPLITTER AND LASER CABLE HOLDER

This section presents the mechanical design for the lens holders, beam splitter and the laser cable holder which are used to construct the laser scanner system.

APPENDIX

C

LUMICS DIODE LASER

This section provides the detail information about the diode laser system use in this dissertation.

User Manual

LuOcean Mini Diode Laser Module

Figure 1: Image of the LuOcean Mini 4(upper) and Mini 8 (lower) Diode Laser Module

Original date of issue: 18.12.2015
Revision date: 03.12.2019

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User Manual

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1 Overview

This document provides information to installation, handling and operation of this product. Lumics GmbH shall not be liable for errors in this manual or damages in connection with the use of this material. The information contained in this manual is subject to change without prior notice. Read this operating instruction manual carefully before putting this product into operation. This product is an OEM product for integration into other equipment and is only to be used as component. The equipment integrator is responsible for CDRH (center for devices and radiological health) or similar certification of all systems sold with this product. He is also responsible for the certification and use of this diode laser according to EN60825-1 and its appendices.

1.1 Revision History

Manual Version	Where	Description
12.10.2018	Chapter 2	Improvements for revision >=4.0 (>=3) of the Mini4 (Mini8). Detailed description of the operation limits of the pilot beam and the fiber sensor. 12 pin signal connector with locking. Maximum torque to fix the fiber to the FSMA port is increased from 0.2Nm to 0.5Nm. Internal circuit to prevent over current to the pilot laser by manual setting of the pilot intensity. Enhance source current of the fiber sensor to 10mA.
23.1.2019	Chapter 2	Wiring of the fiber sensor corrected
03.2019	Chapter 1	Update warranty and safety precautions, re-arrange Chapters
26.8.2019	Chapter 2	Update information to diode laser power cable
16.9.2019	Chapter 3.5	Correct Molex signal connector type
03.12.2019	Chapter 1.5	Add warranty is lost by a broken label

Table 1.1 – Revision History

1.2 General Safety Precautions

The product is a CLASS-I laser product with invisible infrared laser radiation. Avoid any eye or skin exposure to direct or scattered radiation light emitted from the optical fiber output of the diode laser module. Always use protective eye wear during operation of the laser. For guidance please consult user standards such as ANSI, ACGIH, or OSHA. Never open the protective housing of this laser. All the service and maintenance work should be performed by Lumics employees only.

The laser is a source of intense light having characteristics that are very different from the light emitted from conventional light sources. Therefore, before attempting to operate the laser, the user must be aware of these characteristics of laser light and the proper safety precautions. The energy level of the laser beam is high enough to cause serious injury to skin or eye with probable loss of vision if the beam were to pass directly into the eye. When the laser beam is collimated, the energy in the beam remains high and dangerous even at great distances from the laser.

1.3 Safety Precautions when using with a Laser

- Please consider the following safety precautions before operating a laser with this fiber cable
- NEVER LEAVE THE LASER ON, OPEN, AND UNATTENDED!
 - NEVER LOOK DIRECTLY INTO THE MAIN LASER BEAM!
 - Do not allow untrained and not experienced personnel to handle the laser.
 - Never open the diode laser.
 - Disconnect the power supply from mains when not operating the diode laser.
 - Remove the power supply when inspecting or cleaning the fiber connector
 - Always avoid placing reflective objects in the laser beam. Laser light scattered from a reflective surface can be as damaging as the original beam.
 - When the laser is on and the laser output beam is not being terminated in an experiment or optical system, the beam should be blocked and guided into a beam dump.
 - Post warning signs when laser is in operation, limit access to the laser area.

User Safety

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3.2 Mounting on a Heat Sink

Prior to putting this product into operation mount it onto a heat sink. Fix the module with four M3 screws to the heat sink in a X-style fashion with the initial torque set to 0.6 Nm and a final X-style bolt-down at 1.2 Nm. The heat sink must be capable to maintain the operating temperature range of the laser module specified in the datasheet. A heat sink with flatness of 20 microns or less over the entire mating surface to the module and a roughness grade not less than N7 (i.e. Ra=1.6µm) works best.

High thermally-conductive material may be used to eliminate air gaps from the pump module – heat sink interface. In any case even when no thermally-conductive material is used we recommend a thermal resistance between the base and the heat sink of less than 0.1 K/W. Electrically insulating heat films are not recommended and not needed because the laser diode itself is electrically isolated from the package. Ideally, thermally conductive materials are used as an interface between the module and the heat sink to account for any flatness/smoothness discrepancies between the two parts. Suitable thermally conductive materials are listed below.

(1) Thermal foil Manufacturer No EYGS091207 from distributor RS Components with Order No. 712-3911

- Material: Graphite
Thermal Conductivity: 1000W / m · K
Dimension: 0.07mm 115 mm x 90mm.
- (2) Thermal grease
- Example Material: Arctic Silver 5 from Arctic Silver Incorporated from <http://www.arcticsilver.com>
5 also contains sub-micron zinc oxide, aluminum oxide and boron nitride particles. These thermally-enhanced ceramic particles improve the compound's performance and long-term stability.
- Viscosity: Arctic Silver 5 does not contain any silicone. The suspension fluid is a proprietary mixture of advanced, polysynthetic oils that work together to provide three distinctive functional phases. As it comes from the syringe, Arctic Silver 5's consistency is engineered for easy application.
- If you use Peltier elements to spread the heat along the heat sink, please make sure that you use a mounting plate between laser and the Peltier elements (e.g. copper plate of >5 mm thickness).

3.3 Choosing Heat Sinks from Heat load, Wall plug Efficiency and Thermal Resistance

From the product data sheet one can calculate the necessary heat resistance of the external heat sink at operating conditions. The goal is to keep the internal module temperature or the base of the module close to 25°C to maintain the power level given by the data sheet because we have a power drop of approx 5% (7% power per 10K temperature rise at wavelength <1100nm) (above 1300nm). Temperature can be set at a higher level if there is no better external heat sink possible but then optical power and life time at a given current drops. For the design of the external heat sink one needs to know and understand calculation of wall plug efficiency and thermal heat resistance.

The wall plug efficiency (or conversion efficiency) is the ratio of the optical power and the total electrical power. The thermal resistance is defined as the ratio of the temperature increase and the thermal load. The heat flows from the junction of the laser diode through a chain of thermal resistors. These are the laser diode heat sink inside the module, the case of the module, the thermal resistance of heat film and the external heat sink outside the module.

With the definition of wall plug efficiency (or conversion efficiency) = input electrical power / output optical power and thermal resistance = thermal load / total electric power the total power balance of the diode laser module is

$$\text{Thermal resistance} = \text{thermal load} / \text{total electric power} = \text{input electrical power} / \text{output optical power} \text{ and}$$

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the specification of the fiber cable data sheet (see www.lumics.com) and that the fiber connector dimension and tolerance match the requirements given by the data sheet. A bad match may burn the fiber within seconds at full power. The fiber cable connector may become hot during operation due to scattered light to the fiber cladding. Refer to a cable manufacturer for a maximal allowed operation temperature of the cable. Under normal conditions the connector must be able to dissipate about 2%-4% of the output power.

It is worth to measure the temperature of the fiber connector close to the module under maximum power. We recommend that the fiber connector is not hotter than 55°C after 2min of operation at ambient temperature of 25°C.

Before connecting the fiber to the laser use a fiber microscope to inspect its facets. Make sure that the fiber facets are absolutely clean and free of particles, grease or any other contamination. If this is not the case, use a lens cleaning tissue permeated with isopropyl alcohol and carefully clean the fiber facet or use special fiber cleaning tools. FSMA fiber connector can be turned and may suffer from lower coupling efficiency by fiber eccentricity exceeding the specification by the data sheet. This is especially important in case of a 105 µm core fiber. In order to find that out you may set the laser power to a low value (for example 5W). Then rotate the connector by 90° and note maximum deviation. It should be below 5% otherwise the fiber is misaligned.

Important note!

Turn the fiber connector to a tight position by hand or with a wrench using a **maximum torque of 0.2 Nm up to revision <4.0**. Revision >4.0 allows a maximum torque of 0.5 Nm. When the maximum torque is exceeded the internal fiber stop is bent the fiber sensor may be damaged and the fiber end facet is moved away from the optimum internal focus plane.

Maximum torque to fix the FSMA connector (Nm)	LuOcean Mini 4 revision	LuOcean Mini 8 revision
0.2	<4	<3
0.5	>4	>3

Table 3.1 – Maximum torque to fix the FSMA connector to the module

3.5 Electrical Wiring

This product is very sensitive to over current and static discharge. Operate the laser module only in an environment with appropriate grounding and antistatic precautions. Do not exceed the maximal allowed operation current. The laser module is equipped with two different connectors (see fig. 3.1). The connector in fig. 3.1 (left) supplies the current for the laser diodes inside. The connector in figure (right) allows to select the signals from the built in sensors of the laser module and electrically control the build in options.

The power and signal connectors are listed in the fig. 4 below. Please do not solder directly onto the connectors. The signal connector type was improved for different revisions according to the table 2.3 below.

Signal connector type on module side	LuOcean Mini 4 revision	LuOcean Mini 8 revision
3M 159112-5012 without locking	<4	<3
Molex 87833-1231	>4	>3

Table 2.3 – Signal connector type used in different revisions of Mini4 and Mini 8

Fig. 3.1 Power connector (left) and signal connector (right) of LuOcean Mini

3.5.1 Counterparts for diode signal cable

The counterpart of the signal connector is a Molex Milli Grid Cable to Board Receptacle Part No. 87568-1231 with locking ramp. Please use Flat ribbon cable with pitch of 1mm and AWG28.

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3.2 Mounting on a Heat Sink

Prior to putting this product into operation mount it onto a heat sink. Fix the module with four M3 screws to the heat sink in a X-style fashion with the initial torque set to 0.6 Nm and a final X-style bolt-down at 1.2 Nm. The heat sink must be capable to maintain the operating temperature range of the laser module specified in the datasheet. A heat sink with flatness of 20 microns or less over the entire mating surface to the module and a roughness grade not less than N7 (i.e. Ra=1.6µm) works best.

High thermally-conductive material may be used to eliminate air gaps from the pump module – heat sink interface. In any case even when no thermally-conductive material is used we recommend a thermal resistance between the base and the heat sink of less than 0.1 K/W. Electrically insulating heat films are not recommended and not needed because the laser diode itself is electrically isolated from the package. Ideally, thermally conductive materials are used as an interface between the module and the heat sink to account for any flatness/smoothness discrepancies between the two parts. Suitable thermally conductive materials are listed below.

(1) Thermal foil Manufacturer No EYGS091207 from distributor RS Components with Order No. 712-3911

- Material: Graphite
Thermal Conductivity: 1000W / m · K
Dimension: 0.07mm 115 mm x 90mm.
- (2) Thermal grease
- Example Material: Arctic Silver 5 from Arctic Silver Incorporated from <http://www.arcticsilver.com>
5 also contains sub-micron zinc oxide, aluminum oxide and boron nitride particles. These thermally-enhanced ceramic particles improve the compound's performance and long-term stability.
- Viscosity: Arctic Silver 5 does not contain any silicone. The suspension fluid is a proprietary mixture of advanced, polysynthetic oils that work together to provide three distinctive functional phases. As it comes from the syringe, Arctic Silver 5's consistency is engineered for easy application.
- If you use Peltier elements to spread the heat along the heat sink, please make sure that you use a mounting plate between laser and the Peltier elements (e.g. copper plate of >5 mm thickness).

3.3 Choosing Heat Sinks from Heat load, Wall plug Efficiency and Thermal Resistance

From the product data sheet one can calculate the necessary heat resistance of the external heat sink at operating conditions. The goal is to keep the internal module temperature or the base of the module close to 25°C to maintain the power level given by the data sheet because we have a power drop of approx 5% (7% power per 10K temperature rise at wavelength <1100nm) (above 1300nm). Temperature can be set at a higher level if there is no better external heat sink possible but then optical power and life time at a given current drops. For the design of the external heat sink one needs to know and understand calculation of wall plug efficiency and thermal heat resistance.

The wall plug efficiency (or conversion efficiency) is the ratio of the optical power and the total electrical power. The thermal resistance is defined as the ratio of the temperature increase and the thermal load. The heat flows from the junction of the laser diode through a chain of thermal resistors. These are the laser diode heat sink inside the module, the case of the module, the thermal resistance of heat film and the external heat sink outside the module.

With the definition of wall plug efficiency (or conversion efficiency) = input electrical power / output optical power and thermal resistance = thermal load / total electric power the total power balance of the diode laser module is

$$\text{Thermal resistance} = \text{thermal load} / \text{total electric power} = \text{input electrical power} / \text{output optical power} \text{ and}$$

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- Example Material: Arctic Silver 5 from Arctic Silver Incorporated from <http://www.arcticsilver.com>
5 also contains sub-micron zinc oxide, aluminum oxide and boron nitride particles. These thermally-enhanced ceramic particles improve the compound's performance and long-term stability.
- Viscosity: Arctic Silver 5 does not contain any silicone. The suspension fluid is a proprietary mixture of advanced, polysynthetic oils that work together to provide three distinctive functional phases. As it comes from the syringe, Arctic Silver 5's consistency is engineered for easy application.
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3.3.2 Counterparts for the diode laser power cable
For Mini 4 this is a Molex Mega-Fit Receptacle Housing Part No. 171692-0106.
For Mini 8 this is a Molex Mega-Fit Receptacle Housing Part No. 171692-0108.
Please use Molex Mega-Fit Female Crimp Terminal Part No. 76923-0322.

The pinout of the power connector depends on your individual laser module configuration and is shown in the data sheet or the test report of your module. Please note that this power connector can carry max. 26 A per pin.

The connector pins are to be contacted with 3.3 mm² AWG 12 power cables. Ensure that the power supply is switched off before connecting the cables. The (+) and (-) laser diode pin out of the connector is shown in the datasheet/test report for each individual module. Warning: Do not cross the polarity. This can cause irreversible damage to the diode laser.

Note: that the laser diodes are electrically insulated from the laser module case.

Pin	Configuration
1	Vcc (11-13V) to Fiber Sensor (Rev >=4) and Monitor Photo Diode
2	Fiber Sensor 1 Out (0-(Vcc-0.7V)) *
3	GND1
4	Fiber Sensor 2 Out (0-(Vcc-0.7V)) *
5	Monitor Photo Diode 1 Out (0-4(**) V) *
6	Pilot Laser Supply Voltage (red: 5 V, green: 8 V) *
7	Monitor Photo Diode 2 Out (0-4(**) V) *
8	GND2 for Pilot Laser
9	NTC / PT100 or LM35 Supply Voltage +5 V *
10	Pilot Intensity Control In (0-5 V) *
11	NTC / PT100 or LM35 Signal
12	No connection

Table 3.2 Pinout of signal connector (*=optional, ** max of 0.5V only for 19xxnm)

3.6 Pilot laser (Optional green and red)

Connect the pilot laser driver onto pin 6 (+) and pin 8 (ground). The supply for the pilot lasers must not exceed the specified two ways. A schematic view of the internal wiring is shown in Fig. 3.2. The intensity of the pilot laser can be adjusted along with the pilot lasers. At factory the power of the pilot laser is set to a value of approx. 3mW using the manual way (1) or via control voltage (2) of 0V is applied to pin 10 in case the customer orders the module together with a Lumics driver board. The intensity control of the pilot is realised by a constant power control loop which adjusts the pilot current such that the pilot intensity setting remains constant. In the following the two only ways how the pilot intensity can be set are described:

- (1) Manually by the internal trimpot R6 accessible through the hole in the cover (see Fig. 2). In this case the pin 10 must be left open (floating). The trimpot R4 has no effect. At factory the red pilot is set by this trimpot to the standard value of app. 3mV (No Lincics driver board is ordered together with the module. Please use a screwdriver of width 1.5mm to tune the pin R4 and R6).
- (2) Remote by an external voltage applied to pin 10. This voltage is referenced to the GND2 of the pilot laser and must be within the range 0.5V (maximum power of the pilot laser, 0V is about 90% of maximum power) to 15.5V (pilot laser off). In order to control the full power range from off to maximum pilot intensity both trim pots R4 and R6 are set in factory by a tuning procedure described below and can be reset by the customer; if R4 and R6 have been set at the customer's side. The pilot intensity depends on the control voltage and if pin 10 is left floating the pilot intensity value is unknown.

Caution: Never exceed the maximum current and minimum / maximum voltage as below for the following type of pilot laser. This can happen by higher than specified supply voltage to pin 6 or Pin10 or by wrong manual setting of the pilot intensity using the trimpot R4 and R6.

- Red pilot: maximum supply current/voltage 35mA/5V. The minimum supply voltage is $\geq 4V$ to prevent flickering and unstable operation.
- Green pilot: maximum supply current/voltage 125mA/8V. The green pilot can not be tuned down below 10W power. The minimum supply voltage is $\geq 7V$ to prevent flickering and unstable operation.

Pilot maximum current limitation during manual setting by trimpot R6 and R4	LuOcean Mini4 revision	LuOcean Mini8 revision
Red pilot maximum current limitation during manual setting by trimpot R6 and R4	Yes for revision >=4 No for revision <4	No for any revision
Green pilot maximum current limitation during manual setting by trimpot R6 and R4	No for any revision	No for any revision

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User Manual

Table 3.3 – Pilot intensity current limitation used in different revisions of Mini4 and Mini 8

3.6.1 Procedure to tune R4/R6 to a proper value in order to get the full power range from 0mW to 3mW by the external voltage control of the pilot laser

- (1) Apply 0.5V of external voltage to pin 10 and tune R6 to maximum power of the pilot laser.
- (2) Apply 5V of external voltage to pin 10 and tune R4 such that pilot laser is fully off and then add a half turn in the same direction of rotation to catch the real off state.

Fig. 3.2 Wiring of the internal pilot beam (upper). Turning the Trimpot 6 (lower)

3.7 Fiber sensor (Optional)

The fiber sensor is a mechanical switch. A second fiber sensor is optionally available. The sensor should be supplied with 12V (+/-1) V pull-up voltage on the pins 1 (+) and 3 (ground). The same pins provide the voltage supply for the second optionally installed fiber sensor. The signals waves from the sensor one and two are provided on pin 2 and 4, respectively and depend on the signal supply voltage. If the SMA connector is correctly plugged in, the signal (pin 2) is 12 V (and also on pin 4, if the second fiber sensor is present). If the SMA connector is not plugged in, the signal at pin 2 is 0 V (and also on pin 4, if the second fiber sensor is present). The maximum output current at fiber sensor signal is given in table 2.5 depending on the revision of the laser module.

Fig. 6.1 Schematic wiring of the fiber sensor signal inside the module

The source current both fiber sensor and monitor can be up to 25mA. A protection circuit limits the current to 100mA.

Pilot maximum current limitation during manual setting	LuOcean Mini 4 revision	LuOcean Mini 8 revision
1000 A	1000 A	1000 A

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by trimpot R6 and R4	revision >=4	revision >=3
Integrated operational amplifier to source a signal up to 10mA	revision <4	revision <3
No integrated operational amplifier and source current limited to 4mA. Requires Vcc (12V-11V) supply for the amplifier. Maximum signal level is Vcc-0.7V		

Table 3.4 – Maximum source current of the finer sensor signal

3.8 Monitor diode (Optional)

The supply voltage for the two monitor diodes is connected on pin 1 (12V-2V) and pin 3 (GND). The monitor diode is reversed biased such that the monitor cathode is connected to pin 1 and its anode to ground (pin 3) because this is the only way to generate a photo generated diode current which is linear to the laser diode power. The supply voltage is shared with the fiber sensor. The monitor signal is on pin 5 for the first photo diode, and on pin 7 for the optionally installed second photo diode.

Standard signal type of the monitor is a voltage signal which is generated by the monitor photo diode photo current in the range of 10µA to 1mA and a voltage drop to ground over an internal trim potentiometer (resistor see Fig. 7) which can be accessed from outside via a hole in the cover (see Fig. 2). The trim pot is factory set such that at max. rated power according to the data sheet the voltage is about 4 V for up to 1500nm and only 0.5V for 19xxnm. The customer can change this max. signal level by adjusting the trimpot R1 and R2 from outside. There are two trim pots for each monitor diode shown in Fig. 2. In case of the option with monitor diode configured as current signal the internal resistor is not installed and the photo generated current is directly wired to pin 5 or 7.

Fig. 7 Schematic wiring of the monitor signal inside the module

3.9 Temperature sensor Options

- NTC (standard) A 10 kΩ NTC is offered as standard temperature sensor. It is connected between pin 9 and pin 11. The sensor is mounted onto the base plate of the laser module close to the laser diodes. Depending on the base plate temperature the NTC resistance changes in the attachment, chapter 5, page 14.
 - PT100/PT1000 (optional): As alternative to the NTC temperature sensor also temperature sensors PT100 or PT1000 are offered. These are connected between pin 9 and pin 11 similar to the NTC.
 - LM35 (optional): The LM35 is optionally available instead of the NTC. It is calibrated in °Celsius. It provides a 10 mV/°C scale factor and an accuracy of 0.5°C at 25°C. It operates for supply voltages ranging from V_S = 4 V to 20 V. The temperature range fully covers the maximum operating temperature range of the diode laser. Pin 9 (+ V_S) and pin 3 as ground (GND) are used to apply the supply voltage to the LM35. The signal line (V_{out}) is provided on pin 11. The range of V_{out} is between 150mV at 15°C and 350mV at 35°C during operation.
- The Mini-8 diode laser module can be offered with a second temperature sensor on request. If available this is connected to pin 12.

4 Operation

Read the safety precautions before putting the diode laser system into operation. You must also read the manual of your laser diode driver, power supply which is used to operate this product.
Make sure that the output fiber is absolutely clean and free of particles, grease or any other contamination. Make sure that the optional protective window behind the fiber receptacle is absolutely clean. Ensure that no tools other the appropriate fiber connector are plugged into the fiber receptacle which cause scratches on the window and damage the laser module. The receptacle must be covered all the time by a dust cap when the fiber is not plugged into the receptacle.

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Avoid strong back reflection from external elements of more than 10% (30% for diode laser up to 1200nm (>1400nm), originating for example from mirror like metal targets or fiber laser fiber gratings or external laser cavity mirrors. In case the external element acts as second laser cavity for the laser diode the light intensity is amplified by and could reach the threshold damage level of the laser diode (COMD) and could damage the internal laser diodes or could be absorbed within the optical fiber connector. The optical system must be designed to reduce harmful back reflection targeting the output fiber and the fiber connector.
Make sure that the internal temperature control works. Overheating may result in severe damage to the laser diodes and to the fixing of the internal optics (see data sheet for maximum ratings).

4.1 Putting the diode laser into operation

Perform the following steps in order to put the diode laser into operation:

- Put the heat sink and temperature control into operation and ensure that the temperature of the diode laser is within range given by the data sheet
- Remove the dust cap from the optical output connector receptacle
- For pilot option only: The pilot beam may be turned on now by applying voltage to the pins described above
- The diode laser is now ready to be turned on, by applying the operating current to the power connector. See detailed operating current and voltage written in the test report for each individual diode laser to the power connector.
- Now the diode laser is ready for continuous pulsed or c.w. operation.

4.2 Operation notes to prevent damage to the laser diode module

This diode laser operates with passive cooling. Temperature stabilization within the range given by the data sheet should be implemented.

- Make sure that laser and laser diode driver/power supply are connected before switching on laser diode driver/power supply. Never disconnect laser and laser diode driver/power supply before switching off laser diode driver/power supply. Uncontrolled on/off switching may cause current spikes which can result in fatal device damage. This applies also for the pilot laser.
- Laser diodes may be damaged when the maximum forward current exceeds the value given by the test report by 10%.
- Make sure the reverse voltage to the laser diode is below the forward operating voltage noted in the test report. If this reverse voltage exceeds this value the laser diode which is a pn junction device may break through and is irreversibly damaged.
- Laser diodes are ESD sensitive device. Please ensure that laser diode electrical connection is not connected to parts and humans which are electrically loaded.
- The temperature of the laser during operation must be within the specified temperature range to ensure optimal lifetime of the laser. Never operate the laser without or with insufficient cooling.
- Do not leave the fiber connector without protective cap when the diode laser is not in operation, in order to avoid fiber contamination.

4.3 Fiber Connector Handling

Dust, grease and other contaminations of the fiber facet cause light absorption and local overheating. In some cases it can cause a fiber degradation making a laser operation impossible. Operate laser only with clean fiber facet. Therefore:

- Avoid dusty environments.
- Do not touch the fiber front facet and not allow it contacts with other surfaces.
- Periodically control the fiber and clean if necessary.
- Do not expose the fiber to temperature higher than 55°C during storage or operation.
- Do not pull the fiber.
- Do not bend a fiber with a radius smaller than 200x its cladding diameter (rule of thumb).
- Avoid strong back reflection targeting the connector.

Use a clean-room compatible tissue, put some isopropyl alcohol onto it and carefully clean the facet of the fiber facet or connector facet, or use special fiber cleaning tools. Perform cleaning only with power supply switched off! There are many materials commercially available for fiber optic cleaning. Some are marketed specifically for the fiber optics industry. Isopropyl alcohol is the most commonly used one. Isopropyl alcohol will loosen particular contaminants and allows to remove oil or grease from the fiber. It is used on swabs and wipes, by directly spraying, in soaking tubs, and in ultrasonic cleaners.

Please note that for all SMA free standing fiber connectors the fiber tip does not have physical contact to the female. Thus do not apply much force by cleaning. Otherwise you can break the fiber tip.

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4.4 Replacement of Protection Window (optional)

Optionally, your laser module is equipped with a replaceable protection window in the fiber output connector. Perform the following steps to replace the window. The replacement should be carried out in a clean and dry environment. Any dust and moisture has to be avoided.

Figure 4.1: Demonstration how to remove the protection window (picture shows Mini-4)

1. Order the protection window from Lumics.
2. Turn the operating current to zero prior to window replacement. Ensure that there is no electrical current applied to the laser and to the signal connector. In case of uncertainty it is recommended to remove the signal connector.
3. Open the screw on top of the laser module which is close to the fiber output connector using a cross slot screw driver (see left part of above Figure)
4. Carefully pull out the window carrier by using an angled tweezer (see right part of above Figure)
5. Replace the window carrier with the window and put the new window back into the opening as shown in the right part of above Figure.
6. Tighten the mounting screw as shown on the left part of above Figure.

Now the Diode laser is again ready for operation.

5 Trouble shooting

5.1 Laser does not emit

- Check correct fiber cable type
- Check correct connection of electrical wiring or possibility of short circuits (e.g. removal of short circuit bridge).
- Check correct current and voltage settings in your laser diode power supply.

5.2 Output power or other specifications cannot be reached

- Check correct current, voltage, and cooler settings of your laser diode controller.
- Check if optical fiber is bent, damaged or dirty.
- Check if laser protection window (optional) is dirty or damaged.
- Ensure laser module is sufficiently cooled.
- Check monitor diode for correct calibration.

5.3 Laser temperature too high

- Check thermal contact of the laser module to its heat sink.
- Check if heat sink is sufficiently designed
- Check if wavelength is similar to the data delivered (a higher wavelength can be a sign of bad thermal contact).
- Check settings of the temperature sensor, recalibrate it if necessary.

5.4 Fiber sensor (optional) does not detect

- Check if electrical wiring to the fiber sensor signal pins is correct
- Check if FSMA connector specifications and tolerances are correct (look at fiber cable data sheet www.lumics.com)

5.5 Pilot laser power too low

- Check current and voltage supply.

Driver Kit Test Data Sheet

1078266

Serial no and product name of the diode laser which sets the configuration of this driver regarding minimum and maximum operating current and options (pilot, fiber sensor, power and humidity monitor)	Driver kit product name	LU_DR_KB_D_DC_K5-AM
	Driver kit serial no	1078266
	Driver kit firmware	LBA090420V2.2.0
	Date of Test	15. Apr. 20
	Diode Laser Product Name	LU1470D034-E10AN
	Diode Laser Serial No	1064318

Test Parameter	Check	Value / Result	Unit
Controller SN			
1078263			
DC Supply Voltage	set	48	V
RS232 communication 115200 Baud	pass	OK	
Fan Driver start-up / full voltage	pass	12	V
Max. pilot voltage 5V on / off	pass	5	V
Max. pilot voltage 8V on / off			V
Monitor 1 voltage read out	pass	4,1	V
Monitor 2 voltage read out	pass	NA	V
Fiber sensor 1 read out	pass	Error Code not "0"	
Fiber sensor 2 read out			
Temperature Sensor NTC 1	pass	20,1	°C
Temperature Sensor NTC 2			°C
External interlock open	pass	OK	
Turn laser on via RS232 with external interlock open - laser should stay off	pass	OK	
Over internal temperature of laser switches laser current off	pass	OK	
Interlock Signal (Pin 8, T2.1)	pass	OK	
Firmware Update Process working	pass	OK	
Laser driver at logical address 0 SN			
1078264			
DC Supply Voltage	set	24	V
Test max. current setting cw for 1min dedicated to above diode laser	pass	13	A
Test min. current setting cw for 1min dedicated to above diode laser	pass	0,5	A
Test max. current setting (width = 1ms & d.c = 50%) dedicated to above diode laser	pass	13	A
Test min current setting (width = 1ms & d.c = 50%) dedicated to above diode laser	pass	0,5	A
Actual current @ lop cw	pass	12,79	A
Measured current @ lop	pass	12,8	A
Laser on signal out (Pin 2, T2.2)	pass	OK	V
Rise time (pulse 10ms, d.c. 50%, 12,79A)	pass	12	µs
Rise time (pulse 1ms, d.c. 50%, 12,79A)	pass	12	µs
Rise time (pulse 1ms, d.c. 50%, 0,5A)	pass	3	µs
Ripple (cw, I _{max})	pass	<0,15	A
Chiller driver 1 SN			
1078265			
DC Supply Voltage	set	48	V
Maximum current to chiller unit	set	14	A
Chiller Unit temperature limit	set	70	°C
Cooling function	pass	OK	
Heating function	pass	OK	
Cooling and heating regulation	pass	OK	
Over heating protection	pass	OK	

Values are measured with forced air cooling, ambient temperature between 20°C and 30°C and a test diode modul connected by cable of <1m length and Lumics GUI demonstrator software as described in the manual

Precautions

Warning: Only technically qualified personnel should install, maintain and troubleshoot the equipment described herein.

The driver kit does not have redundancy or monitoring of its electrical components. It must be connected to a DC voltage supply ranging from 12V to 48V depending on the diode laser. With proper housing medical EMV and ESD tests can be passed successfully. Special care must be taken to ensure that during handling and operation proper ESD practices are followed.

- (1) Follow the instructions and limits given by the manual and the data sheet to operate the device to prevent damage to the device and loose warranty
- (2) Operating temperature and rel. humidity must be chosen in a way that the dewpoint of humid air is below the temperature of the board to avoid condensing of water.

Test Data Sheet

* 1 0 6 4 3 1 8 *

Laser parameter	Symbol	Value	Unit
Operating power	P_{op}	34.00	W
Peak wavelength	λ_{peak}	1470	nm
Threshold current	I_{th}	1.06	A
Operating current	I_{op}	12.79	A
Operating voltage	U_{op}	11.15	V
Pilot beam wavelength ⁽¹⁾	λ_{pilot}	650 ±10	nm
Measured Features	Symbol	Value	Unit
Pilot laser output power	P_{pilot}	3.0	mW
NTC resistance @ 10p	R_{th}	9911.31	Ω
Operating monitor voltages	$U_{mon,op}$	4.1	V
Fiber sensor check		pass	

Notes: Values are measured at an internal sensor temperature of 25°C and at the fiber end with a 200/220 NA 0.22 multimode fiber. Plot laser and fiber sensor supply voltage is set according to data sheet. (1) Plot laser wavelength is not measured but guaranteed to be in the given wavelength range by design at measured power of 3mW

Power, Voltage, Monitor Characteristics

Spectral Characteristics

Model	LU1470D034-E10AN
Serial No	1064318
Date	4. Mrz. 20

Pinout

Connector - laser diode supply
MolexTM connector (Part No. 172064-0008).
Pin connection dependent on individual electro-optic configuration. Maximum current per pin is 20 A.
Total current to cathode exceeds 20 A. Two pins must be connected to cathode of driver board.

Counterparts for external cable
Molex Mega-Fit Receptacle Housing Part No. 171605-0001
Molex Mega-Fit Female Crimp Terminal Part No. 78823-0022

Connector Signals

™ connector on laser module
Part No. 159112-5012)
Counterpart for external cable
Moles Mill Grid Cable to Board Receptacle
Part No. 87568-1273

PIN	Configuration
1	no connection
2	no connection
3	no connection
4	Laser diode anode (+)
5	Laser diode cathode (-)
6	no connection
7	no connection
8	no connection

Pin	Configuration
1	Supply V12
2	Reset/ Sensor1 Signal
3	GN1
4	Fiber Sensor2 Signal
5	Monitor Photo Diode 1 Out (0-4V)
6	Phot Supply (5V red, 8V green)
7	Monitor Photo Diode 2 Out (0-4V)
8	Phot GN2
9	NTC / PT100 / LM35 Supply 5V
10	Phot intensity control in (0-5V)
11	NTC / PT100 / LM35 Signal

Precautions

- ... The laser light emitted from this device is invisible and may be harmful to the human eye and skin. Avoid looking directly into the fiber when the device is in operation. Use of protective gear is recommended when the device is in operation.
2. **Power supply:** Transient electric spike (above maximum rated current) may cause a damage to the laser. A surge-free power supply and a slow starter circuit should be used. To avoid causing an electrical surge, pins should be connected or disconnected on the fixture before turning the power off.
3. **Electrostatic discharge is the primary cause of unexpected laser diode failure.** Take extreme precaution to prevent ESD. Use wrist straps, grounded work surfaces and rigorous antistatic techniques when handling this device.
4. **Extremely careful attention is required when this pump laser module is in operation.** This device should be operated and handled by experienced personnel. Any doubted matters during or before operation should be discussed with Lumics.
5. **This product is not certified by with IEC 60825-1 or 21CFR1040.101 and must comply with the applicable regulations by the Purchaser if sold as laser product.**

User Safety

LU_LWL0200_0280/0500_220A2A1

Test Data Sheet

Model	LU_LWL0200_0280/0500_220A2A1
Serial No	1065584
Fiber Serial No	20050054
Date	16. Apr. 20

Fiber parameter	Value	Unit
Core diameter	200	µm
Cladding diameter	280	µm
Numerical aperture	NA 0.22	
Fiber termination type	SMA905	
		mm
		µm
		µm
Visual inspection by operator	pass	mW

Fiber Input

Fiber Output

APPENDIX

D

POSITION SENSING DETECTOR

This section presents the data sheet of the PDP90A lateral position sensor that is used to measure the output position of the fiber optic and is applied to the designed controller to improve the performance of the constructed system.

PDP90A Position Sensing Detector

User Guide

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Chapter 1 Warning Symbol Definitions

Below is a list of warning symbols you may encounter in this manual or on your device.

Symbol	Description
Direct Current
Alternating Current
Both Direct and Alternating Current
Earth Ground Terminal
Protective Conductor Terminal
Frame or Chassis Terminal
Equipotentiality
On (Supply)
Off (Supply)
In Position of a Bi-Stable Push Control
Out Position of a Bi-Stable Push Control
Caution: Risk of Electric Shock
Caution: Hot Surface
Caution: Risk of Danger
Warning: Laser Radiation
Caution: Spinning Blades May Cause Harm

Chapter 2 Description

The PDP90A is a two-dimensional, tetra-lateral position sensing detector (PSD) that provides X and Y-axis positional information. It is designed to work with Thorlabs' TPA101 T-Cube Auto-Aligner. This sensor can be used for general alignment purposes, as well as precise position measurements. This sensor gives the position of any spot contained within the active region and supports spot sizes up to 9 mm in diameter. A spot between $\varnothing 0.2$ mm and $\varnothing 7$ mm is recommended.

Thorlabs also manufactures the PDQ80A quadrant position detector for precision alignment needs.

2.1. Setup Instructions

- a. Unpack and mount the PDP90A position-sensing detector. An AS4M8E M4 to 8-32 thread adapter is included for customers using metric opto-mechanics.
- b. Connect the sensor to a TPA101 T-Cube Auto-Aligner.
- c. Follow the directions in the controller's operating manual to ensure that the position-sensing detector is properly communicating with the controller.
- d. Send a beam onto the active area for measurement. The spot size input beam should be between $\varnothing 0.2$ and $\varnothing 7$ mm. For best results, the spot should be located within the center 80% of the surface of the detector (a 7.2 mm x 7.2 mm square). Adjust the power level so that the SUM output voltage is ≤ 4 V. This ensures the best signal-to-noise ratio and avoids signal saturation.

2.2. Details of Operation

This section of the manual discusses the operating principles of 2D lateral effect position sensing detectors and includes key equations for understanding their output signals.

2.2.1. Overview

2D lateral effect sensors accurately measure displacement - movements, distances, or angles – providing feedback from systems that require precise alignment, such as steering mirrors, microscope focusing, and fiber launch systems. These sensors are light-controlled variable resistors that work by proportionally distributing photocurrent to determine position in the X and Y from the four electrodes when incident light is applied to its surface.

In addition, the PDP90A is a continuous position-sensing detector, which derives position by dividing photo-generated electrons within their substrate, rather than by profiling the intensity distribution on the surface as with segmented detectors. There are two different types of lateral effect sensors available: duo-lateral and tetra-lateral.

Duo-lateral sensors provide a resistive layer on both the anode and cathode photodiode connections. This isolates the X and Y positional information of the sensor and allows the sensor to be highly linear and very accurate. However, the added resistive layers significantly increase manufacturing cost.

Tetra-lateral sensors use a single resistive layer with a common cathode and an anode on each side of the detection area. This makes them very inexpensive to manufacture, but their linearity decreases as the spot moves away from the center. This is caused by the physical location of the anodes along the sides of the sensor, specifically in the corners where the anodes approach each other.

Thorlabs PDP90A detector uses a variation of the tetra-lateral-type sensor commonly called a “pin cushion” position-sensing device. This sensor moves the anodes to the four corners of the sensors and reshapes the sensor area, to produce linearity comparable to a duo lateral sensor at a significantly lower cost.

The tetra-lateral-type, “pin cushion” has a characteristic form of which we will use some important values to determine further calculations. The active area is 9x9 mm at the waist of the pincushion; however, at the outer most regions the area goes to 10mm, therefore the calculations are performed from the widest part of the sensor.

2.2.2. Calculating the Position

The PDP90A includes circuitry to decode the D_x , D_y , and SUM signals as follows (see *Figure 1*):

$$\Delta x = (B + C) - (A + D)$$

$$\Delta y = (A + B) - (C + D)$$

$$SUM = (A + B + C + D)$$

Note: A, B, C and D are four electrodes on the upper surface, formed along each of the four edges

The following formulas will help you to determine the actual light beam position along the X and Y-axis.

$$x = \frac{L_x(\Delta x)}{2 \text{ SUM}} ; y = \frac{L_y(\Delta y)}{2 \text{ SUM}}$$

Figure 1 Layout of a Lateral Effect Position Sensor

2.2.3. Position Resolution

Position resolution is the minimum detectable displacement of a light spot incident on the PSD. Position resolution (ΔR) is a factor of both the resistance length (L_x or L_y), or the X and Y length, and the signal-to-noise ratio (S/N). The signal to noise ratio for this system can be defined as the SUM output signal level (V_0) divided by the output voltage noise (e_n). The PDP90A output noise is $<2 \text{ mV}_{pp}$, or $<300 \text{ } \mu\text{V}_{rms}$:

$$\Delta R = L_x \left(\frac{e_n}{V_0} \right)$$

where ΔR is the resolution, L_x is the detector length (10 mm), e_n is the output noise voltage (300 μV_{rms}), and V_0 is the SUM output voltage level (4 V).

$$\Delta R = 10 \text{ mm} \left(\frac{300 \text{ } \mu\text{V}_{rms}}{4 \text{ V}} \right) = 0.75 \text{ } \mu\text{m}$$

For best results, V_0 should be maximized at 4 V, resulting in a position resolution of 0.75 μm . To do this, monitor the SUM output voltage of the sensor and adjust the optical intensity until the output is approximately 4 V. Outputs above 4 V will saturate the system, causing unreliable readings. The software will indicate the voltage level on a slide bar, and is available for download online at thorlabs.com. To reach the download page go to Services, Downloads, Detectors, and then Position Sensing Detectors and click on the link. If the SUM is saturated, the slide bar color will turn red. Reduce the intensity until it switches to green. This will be equivalent to a 4 V SUM output.

Low power levels should be used for the position sensor. For the PDP90A, the max photocurrent is 40 μA . Using the photosensitivity curve below (*Figure 2*), the system saturation power can be calculated as:

$$P_{max} = \frac{40 \mu\text{A}}{\text{Photosensitivity (in } \frac{\text{A}}{\text{W}})}$$

The TPA101 T-Cube PSD Auto Aligner has a sample resolution of 16-bits with an input voltage range of $\pm 15 \text{ V}$ (30 V range), or a voltage resolution of:

$$V_{step} = \frac{30 \text{ V}}{2^{16}} = 0.458 \text{ mV}$$

$$I_{step} = \frac{V_{step}}{\text{Transimpedance Gain}} = \frac{V_{step}}{100 \frac{\text{kV}}{\text{A}}} =$$

From this, the minimum power required can be determined based on the required accuracy. For example, if the user requires an accuracy of 1%, the minimum photocurrent is $I_{step} \times 100 = 0.458 \mu\text{A}$. Use the formula below to find the minimum optical power for a given wavelength.

$$P_{min} = \frac{0.458 \mu\text{A}}{\text{Photosensitivity (in } \frac{\text{A}}{\text{W}})}$$

Figure 2 Photosensitivity of PDP90A

Chapter 3 Specifications

Electrical Specifications	
Wavelength Range	320 – 1100 nm
Peak Responsivity	0.6 A/W at 960 nm
Transimpedance Gain	100 kV/A
Photocurrent (Max)	40 μ A
Output Voltage Range: Pins 1,2 Pin 3	-4 to +4 V 0 to 4 V
Signal Output Offset	0.3 mV (Typical) 7 mV (Max)
Bandwidth	15 kHz
Spot Size ¹	Ø0.2 to Ø7 mm (Recommended) Ø9 mm (Max)
Power Requirements	±5 VDC ± 5%, 35 mA

Physical Specifications	
Sensor Size	9 mm x 9 mm (0.35" x 0.35")
Mechanical Aperture	Ø0.50" (Ø12.7 mm)
Aperture Thread	Internal SM05 (0.535" - 40)
Dimensions	2.00" x 1.20" x 0.65" (50.8 mm x 30.5 mm x 16.5 mm)
Mounting Thread	8-32, 0.25" Thread Depth (Min)
Included Metric Adapter	External 8-32 to Internal M4 (Thorlabs Item # AS4M8E)
Cable Length	5' (1.5 m) (Typ.)
Connector Plug	Hirose HR10A-7P-6P
Mating Receptacle	Hirose HR10A-7R-6S
Weight	0.25 lbs. (114 g)
Sensor Depth	7.07 mm (0.28")
Operating Temperature	10 to 40 °C
Storage Temperature	-20 to 80 °C

¹ Ø9 mm (Max) at the waist of the pincushion.

Chapter 4 Drawing

Pin	Description
1	X-Axis $[B + C] - [A + D]$
2	Y-Axis $[A + B] - [C + D]$
3	SUM $[A + B + C + D]$
4	+Vsupply (+5 V)
5	Common
6	-Vsupply (-5 V)