

Development of An In Vivo Robotic Camera for Dexterous Manipulation and Clear Imaging

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Degree

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I dedicate this dissertation to my wife, Shafagh Eslami, for her remarkable patience and unwavering love and support over the course of my research.

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Abstract

Minimally invasive surgery (MIS) techniques are becoming more popular as replacements for traditional open surgeries. These methods benefit patients with lowering blood loss and post-operative pain, reducing recovery period and hospital stay time, decreasing surgical area scarring and cosmetic issues, and lessening the treatment costs, hence greater patient satisfaction would be earned. Manipulating surgical instruments from outside of abdomen and performing surgery needs precise hand-eye coordination which is provided by insertable cameras. The traditional MIS insertable cameras suffer from port complexity and reduced manipulation dexterity, which leads to defection in Hand-eye coordination and surgical flow. Fully insertable robotic camera systems emerged as a promising solution in MIS. Implementing robotic camera systems faces multiple challenges in fixation, manipulation, orientation control, tool-tissue interaction, in vivo illumination and clear imaging.

In this dissertation a novel actuation and control mechanism is developed and validated for an insertable laparoscopic camera. This design uses permanent magnets and coils as force/torque generators in an external control unit to manipulate an in vivo camera capsule. The motorless design of this capsule reduces the weight, size and power consumption of the driven unit. In order to guarantee the smooth motion of the camera inside the abdominal cavity, an interaction force control method was proposed and validated.

Optimizing the system's design, through minimizing the control unit size and power consumption and extending maneuverability of insertable camera, was achieved by a novel transformable design, which uses a single permanent magnet in the control unit. The camera robot uses a permanent magnet as fixation and translation unit, and two embedded motor for tilt motion actuation, as well as illumination actuation. Transformable design provides superior imaging quality through an optimized illumination unit and a cleaning module.

The illumination module uses freeform optical lenses to control light beams from the LEDs to achieve optimized illumination over surgical zone. The cleaning module prevents lens contamination through a pump actuated debris prevention system, while mechanically wipes the lens in case of contamination. The performance of transformable design and its modules have been assessed experimentally.

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Chapter 1

Introduction

1.1 Minimally Invasive Surgery

Minimally Invasive Surgery (MIS) is a modern surgical technique, replacing conventional open surgeries, to reduce the patients' operative and post-operative hardships. In comparison with open surgeries, MIS excels in lessening the bleeding, post-operative pain, risk of infection, scarring and hospital stay while enhances the cosmetic results and recovery time. During a multi-port MIS, the surgeon first creates multiple, three to six, relatively small incisions over patients anatomy, mainly in abdominal region. These incisions are usually between 5 mm to 15 mm and are used for insertion of surgical instruments and laparoscopic camera. The surgeons conduct the complex operational tasks by manipulating the surgical instruments, such as grippers, scissors, forceps, suturing needles and etc from outside the patient's anatomy [12]. Performing dexterous surgical tasks requires accurate hand-eye coordination for the surgeons, which is fulfilled by the laparoscopic camera. These cameras are conventionally inserted by pushing long sticks through a separate incision to provide live feedback for the surgeons to control and perform the operational tasks as shown in the left figure¹ of Fig. 1.1.

Considering the kinematic constraints caused by the trocar, manipulation of these rigid stick devices suffers from the counter-intuitive fulcrum effect and the confined workspace. However, expert surgeons have been able to master performing MLS by intensive training

¹<http://angrybirdsriogame.info/?d=Colon+amp+Rectal+Surgery+Associates>

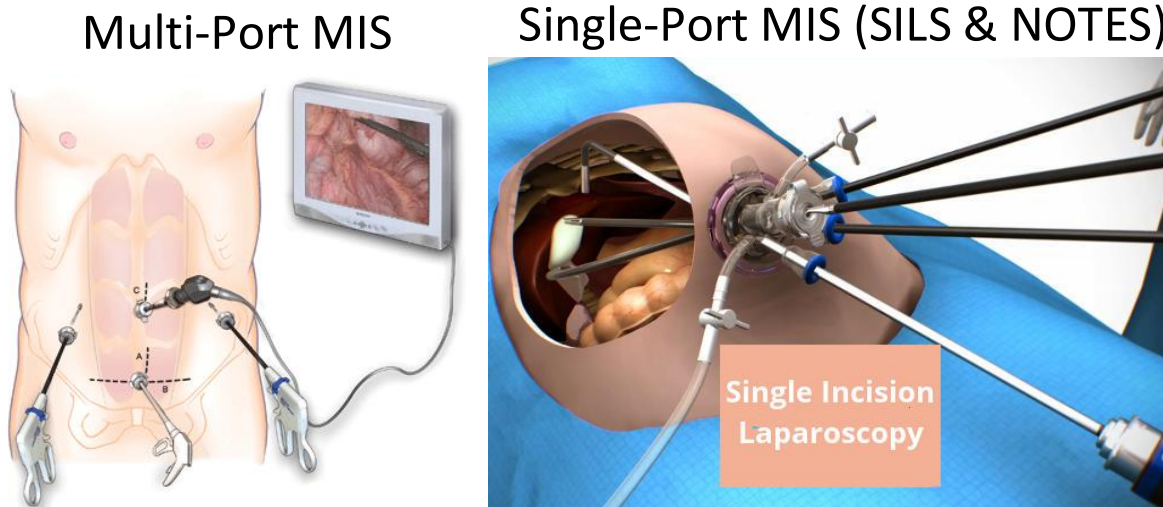


Figure 1.1: Multi-port and single-port minimally invasive surgeries techniques.

and continuous practice and even successfully established MLS as the gold standard for many abdominal surgical procedures such as cholecystectomy, splenectomy, and appendectomy etc [83, 90].

Improvement of surgical techniques led to further reducing the number of incisions by introducing single port laparoscopic surgery (SILS), in which surgeon cut only one incision for insertion, and natural orifice transluminal endoscopic surgery (NOTES), which totally eliminates the incision and use patient's natural orifices for insertion such as umbilicus, mouth and etc.

In SILS and NOTES as shown in the right figure² of Fig. 1.1, all the surgical instruments and laparoscopic camera rod are inserted from a common incision [64, 50]. These recent techniques, further decrease the scarring and bleeding while enhancing the cosmetic results compared to multi-port MIS [15, 63, 81]. Fig. 1.2 schematically shows the enhancement of cosmetic results in different minimally invasive surgeries. Despite the aforementioned advantages, the shared incision used for surgical instruments in SILS and NOTES techniques, deteriorates the control, triangulation and manipulation of instruments and reduces surgeon's dexterity. It also limits the surgical site's view angle for the inserted laparoscopic camera, which may lead to further deterioration of surgical operation. Consequently, these methods

²<https://drheidarizadi.com/learning-center/single-port-surgery.html>

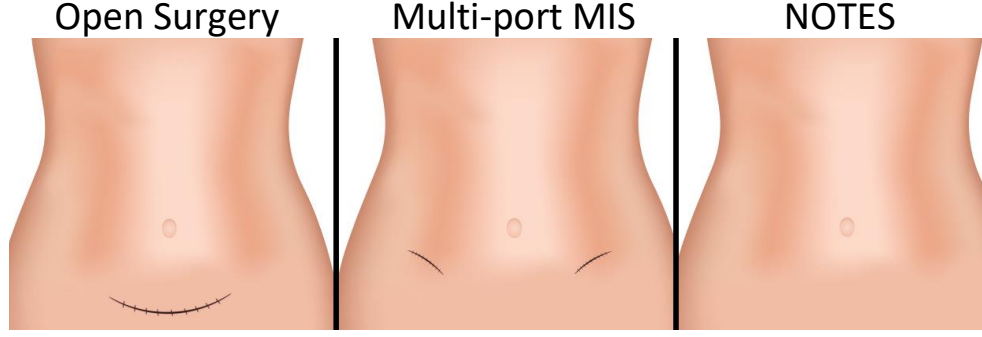


Figure 1.2: Comparison of scarring and cosmetic results in different minimally invasive surgery techniques.

introduce new operational challenges for the surgical team and reduce the ease of operation for many experienced surgeons [Cadeddu and Gahan].

Robotic surgery was emerged as a solution to the traditional laparoscopic surgery challenges. Surgical robotic systems were used in the operation room to mimic surgeons hand motions and perform surgical tasks. *da Vinci*[®] series surgical systems (Intuitive Surgical, Inc., Approved by FDA in 2000) [17, dav] and the Senhance Surgical system (TransEnterix Surgical, Inc., Approved by FDA in 2017), are the leading companies in robotic-assisted minimally invasive surgery (RMIS), which provide unprecedented operational intuitiveness and precision. However, these systems still suffer from limited workspace and kinematic restrictions, specifically in insertion ports. As a result, a second cut becomes inevitable for repositioning these laparoscopic devices so as to get a preferred view angle or better operation triangulation, which may convert a SILS to an MLS or even a OS [29].

Fully insertable robotic camera systems emerged as a promising solution in MIS, where a miniaturized robotic surgical devices in inserted completely into the patient’s abdominal cavity [Cadeddu and Gahan, 80, 60, 93, 54] to provide imaging assistance or even perform possible diagnostic and operative tasks. Implementing robotic camera systems faces multiple challenges in fixation, manipulation, orientation control, tool-tissue interaction, in vivo illumination and clear imaging

1.2 Challenges

In vivo robotic camera systems are proposed as replacement for the camera system in surgical robots, which are costly, bulky and placed ex-vivo. Although in vivo miniature systems enhances the dexterity and maneuverability features, their design and development introduces unique engineering challenges. The challenges mainly emerges in mechanical design, application and clinical acceptance.

1.2.1 Actuation and Mechanical Design

The fully insertable laparoscopic device needs to be effectively anchored and actuated to support designed surgical operations. No physical actuation linkage could be allowed from the outside so that the insertable device could maneuver exibly in the abdominal cavity. Thus, the robot has to either make use of its onboard actuators or seek for a non-contact transabdominal actuation technology. Non-contact transabdominal actuation based on magnetic coupling could be an innovative approach, however it is quite challenging to model and control. To provide reliable manipulation of the camera by the external rotation magnetic field from the stator, the magnetic coupling between the rotor and stator should be capable of generating sufficient force and torque for the translation and rotation of the camera. Compared with a spherical actuator, the air gap between the rotor and the stator is much larger in the surgical situation due to a patient's abdominal wall thickness, which normally ranges is 20 mm $\sim$ 40 mm [70]. The magnetic force and torque will rapidly reduce while the distance from the rotor to the stator increases. Therefore, the design and analysis of the magnetic driving unit and the externally positioned stator will be thoroughly studied in this work for reliable actuation of the camera.

On the other hand, onboard actuators provide improved degrees of freedom compared to magnetic actuation. The main challenge rising in onboard actuators is that they often lead to bulky designs. An innovative design with micro motors has been proposed to overcome this issue and maintain the size within SILS clinical incision limits. The proposed solution uses magnetic actuation for translation, while uses onboard motors to provide extra degrees of freedom for controlling the camera's view angle and illumination.

1.2.2 Motion Control and Tool-Tissue Interaction

Once the laparoscopic camera is inserted into the body cavity, the camera is first manipulated to focus on the operative area. At this stage, we assume that the attractive force between the rotor and the stator is strong enough such that the rotor is pushed against the abdominal wall. The camera system can also be controlled such that the camera is floating in the gas filled body cavity, which requires accurate compensation of gravity by the external magnetic field and estimation of the camera locations. It becomes difficult for the camera system with limited sensing capability. In this work, a contact based control model will be adopted.

The tissue pressure on the rotor is a result of the balanced gravity and magnetic attractive forces to the rotor. The membrane forces are determined by a viscoelastic model consisting of the tissue stiffness and the viscous damping. Considering the variation in the thickness of the abdominal wall, the external magnetic forces should balance the camera gravity but cause little undesired internal pressure to the surrounding tissue. The membrane force is associated with deflection of the tissue, which is an exponential function of the depth of the deflection. With the design of the rotor, the membrane force can be integrated over the depth of the deflection.

In this work, the control model of the camera will be developed according to the force and torque analysis between the camera body and an abdominal wall tissue, in order to realize automatic motion control of the camera.

1.2.3 Optimized Illumination for Superior Imaging Quality

Although in vivo laparoscopic cameras feature the advantages mentioned before, they are still in their infancy for real MIS tasks. One of the major issues that impedes the in vivo cameras from being practical is their inferior imaging performance due to three main reasons. First, the sizes of imaging sensors and optical lenses applied in in vivo cameras are limited by the compact dimensions of the cameras. Second, lighting systems play a crucial role in determining the quality of surgical images beyond the imaging sensors themselves. The state-of-the-art in vivo laparoscopic cameras employ bare LEDs or LEDs combined with poorly designed reflectors. The uncontrolled light beams therefore either waste most of their

energy to illuminate areas outside the cameras FOV or result in bright center and dark margins in the plane of an imaging sensor. In contrast, the conventional laparoscopic video systems introduce external xenon/LED light sources into abdominal cavities via fiber-optics inside the rods. Illumination optical designs are usually only applied to the rods to improve energy efficiency and illuminance uniformity.

1.2.4 Clear Visualization for In-vivo Camera Systems

A noteworthy drawback of in-vivo robotic camera systems is clear visualization during the MIS operations. The camera lens can get dirty with intra-abdominal fluids, blood, surgical rinsing fluid, operation dust particles and smoke which results in image impairment. Researches have shown that approximately one third of the minimally invasive surgeries are performed without clear visualization for surgeons. However, to the best knowledge of authors, None of the robotic systems have considered a prohibitive and/or cleaning mechanism in case of lens contamination. Image impairment reduces the hand-eye coordination and surgeon's dexterity as a result. Also it causes the interruption in the surgery work flow which influences the performance of surgical team.

1.3 State of the Art

Due to the limited surgical spaces inside human bodies, miniature laparoscopy and endoscopy surgical robots with various functions were developed to inspect abdominal cavities, and travel along GI tracts [78, 49], or be manipulated in fluid-filled lumens and/or soft tissues [51].

1.3.1 Laparoscopic surgical camera robots

Insertable imaging robots with magnetic fixation and positioning for laparoscopic procedures have been reported in [18, 74, 5, 67]. In these solutions, the purposes of the on-board magnetic elements are intended for fixation, and manipulation of the device for positioning and orientation adjustments is normally achieved by manually maneuvering an external

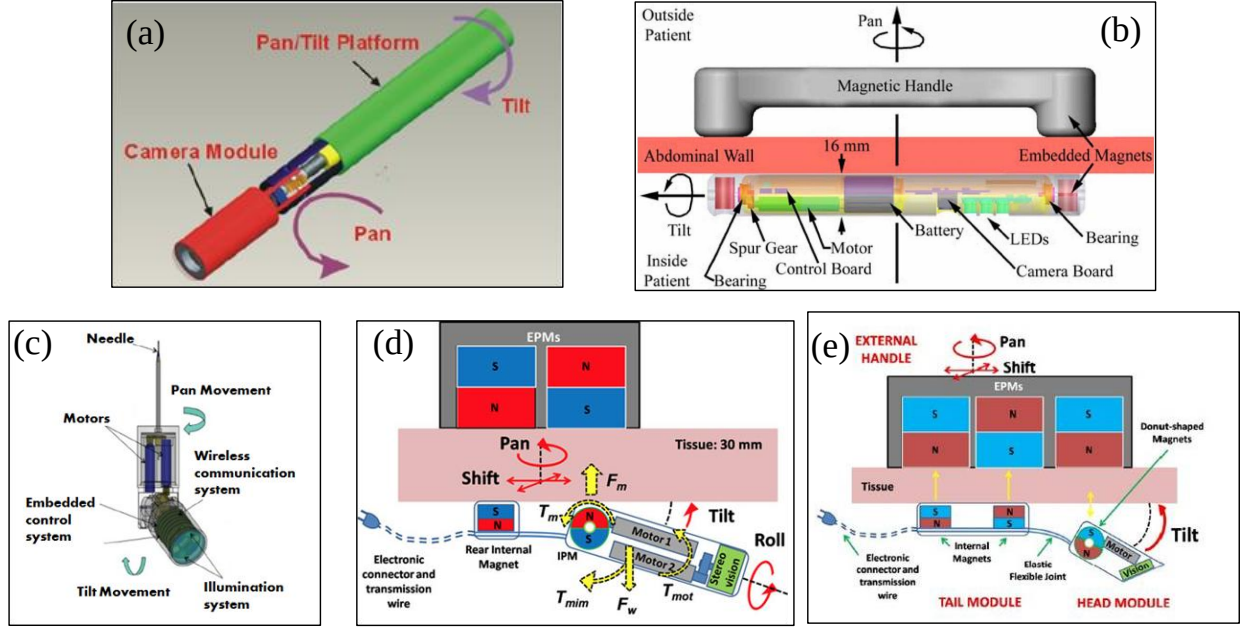


Figure 1.3: Laparoscopic cameras robots. (a) [34]; (b) [56]; (c) [9]; (d) [69]; (e) [68].

permanent magnet. To achieve greater accuracy and controllability of the imaging robots, researches have been done to manipulate the external permanent magnets with precisely controlled robot manipulators to overcome the exponential variability of magnetic fields. Research has also been done to integrate magnetic or electrical driven mechanism into the camera to manipulate the camera components [56, 69, 68], as shown in Fig. 1.3(b), (d), and (e). The existing internal driven mechanisms usually consist of two articulated components, one providing fixation with the abdominal wall, and the other enabling manipulation of the camera module [34, 9]. Tethered multi-link robotic laparoscopic cameras as shown in Fig. 1.3(a) were proposed by [34] which adopt on-board motors and peripheral mechanisms to actuate pan/tilt motion with camera bodies sutured against an abdominal wall for fixation. The camera design proposed in [9] applied a wirelessly controlled motor-driven mechanism for pan/tilt motion with an on-board needle pierced through an abdominal wall for the camera fixation and electronics powering, as shown in Fig. 1.3(c). This articulated structure inevitably increases the size and complexity of the modules.

In spite of the achieved progresses, their inferior imaging quality is a major issue that obstructs the *in-vivo* camera designs to be clinically employed. It is not a straightforward task to obtain high quality images by only improving *in-vivo* imaging systems with HD imaging

sensors. Because for the same sensor size, the higher resolution results in the smaller pixel size that is usually with the lower light sensitivity [20]. Sufficient light intensity within a camera’s FOV is a necessity to enable HD imaging sensors to work in normal condition. In addition, non-uniform lighting distribution on a target visualized area results in poor quality images with shining centers and dim margins. Furthermore, light rays that emit from the on-board light sources should be projected within a camera’s FOV to maximally utilize light energy. The above requirements provide guidelines for designing an optimized illumination system in an *in-vivo* laparoscopic camera.

The illumination systems that are integrated in the state-of-the-art *in-vivo* laparoscopic cameras utilize bare LEDs [69, 34, 5] or LEDs stacked with reflectors [9] to realize basic illumination for the imaging sensors.

A noteworthy drawback of *In Vivo* robotic camera systems is clear visualization during the MIS operations. The inserted camera lens is vulnerable to get dirty often with intra-abdominal fluids, blood, surgical rinsing fluid, operation dust particles and smoke which results in image impairment [8]. Researches have shown that approximately one third of the minimally invasive surgeries are performed without clear visualization for surgeons [92]. However, to the best knowledge of authors, None of the robotic systems have considered a prohibitive and/or cleaning mechanism in case of lens contamination. Image impairment reduces the hand-eye coordination and surgeon’s dexterity as a result. Also it causes the interruption in the surgery work flow which influences the performance of surgical team. In the following we will elaborate the detailed state of the art for laparoscopic lens cleaning systems.

Condensation on the lens frequently happens due to differences between room and intra-abdominal temperatures. Despite using anti-condensation solutions, it can not be solved entirely [85]. Removal of lens will cause a secondary issue and that is the port contamination, which requires extra time for cleaning per each lens removal [77]. In rare cases, some patients are allergic to povidone-iodine solution used for lens defogging when it contacts the Intra-abdominal [62].

Some researchers have addressed these issues by providing *In Vivo* cleaning for conventional laparoscopic surgeries. EndoClear is a device that can be attached to the

internal abdominal wall in the beginning of surgery and be used as a cleaning station during surgery [8]. Disadvantages of this system are: 1) it needs fixation inside the cavity which probably done by suturing which is harmful; 2) the EndoClear itself can get dirty during surgery due to the probable bleeding and fluids inside the abdominal cavity.

An Irrigation system has been developed to wash the lens during surgery [47]. The drawback of irrigation system is that it needs a fluid tube to be inserted along the trocar or an overtube which influences the single port dexterity by increasing port conflict, and may eventually cause larger incision sizes on patients body.

Floshield is a product which uses a plastic overtube that fits over conventional laparoscopes and blows CO₂ and fluid solution over the lens of the laparoscope [7]. Floshield itself increases the trocar outer diameter through the entry port which reduces the dexterity, also it needs external fluid pumping through tubes.

A mechanical wiper has been implemented to clear the lens for *In Vivo* visual robot system with digital defogging by Feng et al [21]. The drawback of this design is the complexity of mechanisms, transferred from motor to actuate the wiper. Additionally, in this system the camera lens is easily exposed to any contamination source and no prevention system has been considered for this purpose.

Digital defogging system has also been used to improve the imaging quality [30]. Although this is an interesting approach, it won't be effective in other contamination types.

1.4 Objectives

The work in this dissertation aims to develop an in vivo robotic camera systems with dexterous manipulation and providing clear imaging during laparoscopy surgery. Specifically, the following four objectives have been targeted.

The **first objective** is to design an automatic rotational control system for a insertable laparoscopic camera capsule with less than 1° control accuracy to provide appropriate vision for the surgeons. This objective is reached by implementing a closed loop control system over a magnetically controlled insertable camera system.

The **second objective** is to recover the lost camera-tissue contact force measurement and characterize the camera-tissue interaction process. This contact force needs to be maintained in an appropriate safe range so that the camera will neither fall off due to insufficient anchoring or damage the tissue due to overload. Meanwhile, the camera-tissue interaction model could relate the contact force with pressure distribution over the contact profile.

The **third objective** is to propose a solution to develop an *in-vivo* laparoscopic camera with an optimized illumination system. this chapter presents an initial prototype of an *in-vivo* robotic laparoscopic camera that features optimized illumination to address the problems, i.e. inferior lighting uniformity and low optical efficiency, in the state-of-the-art designs of *in-vivo* laparoscopic cameras.

The **forth objective** is to propose a solution for frequent lens contamination and image impairing happens during laparoscopic surgery. This chapter presents an in vivo debris prevention system to deviate the probable lens contaminating particles and the wiping mechanism cleans the contaminated lens.

1.5 Contributions

(1) an innovative transformable robotic laparoscopic camera design, which combines compactness for fitting in the abdominal incision and sufficient on-board space for containing the optimized illumination system; (2) an effective freeform non-imaging lens design method for extended light sources, which is used for developing the optimized illumination system; (3) an experimental performance validation of an assembled illumination system with manufactured freeform lenses; and (4) a surgical task demonstration by a complete initial prototype that consists of an *in-vivo* robotic laparoscopic camera, an external anchoring/control unit, and a user controller for sending control signals.

The major contributions of my research presented in this dissertation are listed as follows.

1. An innovative magnetic actuated insertable robotic camera system was developed for SILS with a line-arranged rotor and a hybrid stator as the final design. The design features a reliable unified fixation, translation and rotation control of the

capsulated dummy laparoscopic camera. An closed-loop control system was designed and implemented which can automatically actuate the orientation of the camera with less than 1° control accuracy under an abdominal wall with a normal range thickness.

2. A novel abdominal wall thickness sensing system was proposed and implemented inside the hybrid stator. The sensing system, which consists of four-group tri-axis hall effect sensors, can provide sub-millimeter sensing accuracy in real time. A force sensor is implemented in the camera system which can measure the interaction force online. With sensed abdominal wall thicknesses, and contact force, a control system adjusts the stator-rotor distance in order to prevent the tissue damage during the translation and rotation.
3. An initial prototype of a transformable *in-vivo* robotic camera system integrated with an optimized illumination system for single-port laparoscopic surgery is presented. This work introduced freeform optical lenses into the *in-vivo* robotic camera for controlling light beams from the LEDs to achieve optimized illumination. The designed illumination system features proper illuminance uniformity, optical efficiency, and illuminance on the target area when the camera-to-target distance ranges from 50 mm to 100 mm.
4. A novel *In Vivo* gas generated debris prevention system is developed in order to provide continuous clear vision during surgery. The twofold functions of the proposed cleaning system are: Firstly, it deviates the approaching impairment sources toward the camera lens; Secondly, if the image is impaired by contamination, a mechanical system would be actuated to retract the LED wings, which guides the generated gas to ward the camera lens through a designed edge on a wing. This stream helps to wipe the image impairment cause off the lens surface.

1.6 Dissertation Outline

The dissertation is organized as follows:

Chapter 2 presents the design and control of an insertable laparoscopic camera for minimally invasive surgery.

Chapter 3 elaborates camera-tissue interaction model and study.

Chapter 4 provides detailed design and implementation of an optimized in vivo illumination system.

Chapter 5 discusses the design of debris prevention and lens cleaning systems for maintaining clear vision during surgery.

Chapter 6 concludes the dissertation and shares vision into the future.

Chapter 2

Design and Control of An Insertable Robotic Camera System for Single Incision Laparoscopic Surgery

2.1 Abstract

This chapter demonstrates an insertable magnetic actuated robotic camera system with two-degree-of-freedom (2-DoF) orientation control for single incision laparoscopic surgery. The camera system design consists of an external magnetic control unit and a fully insertable camera capsule. This system features a unified mechanism for anchoring, navigating, and rotating the insertable camera capsule by externally generated rotational magnetic field from the control unit. The motor-free camera capsule is encapsulated in an one-piece housing with two ring-shaped tail-end magnets and one cylindrical central magnet. The control unit which positioned externally consists of both permanent magnets and electromagnetic coils to generate rotational magnetic field and control the camera capsule. An automatic fine orientation control system was developed for the camera capsule. The experimental investigations indicated that our camera control system can achieve less than 1° control accuracies with average errors 0.594° and 0.524° for tilt motion and pan motion respectively.

2.2 Introduction

Minimally Invasive Surgery (MIS) has been widely applied to achieve less operative bleeding and post-operative pain, faster recovery time, and better cosmetic results [15, 63]. This surgical procedure requires making small incisions on patient anatomy to insert surgical instruments, which are manipulated inside patient body. Since the line-of-sight between surgeon eyes and surgical instruments is occluded by patient body, a long-stick laparoscopic camera for visual feedback can be applied through a dedicated incision or a shared incision with other surgical instruments. To improve triangulation and manipulating flexibility between a long-stick camera and other surgical instruments, insertable laparoscopic cameras were developed [74, 5].

Robotic degrees of freedom (DoFs) for an insertable laparoscopic camera are required to adjust the camera's orientation and position inside an abdominal cavity. The cameras need to be mounted on the internal surface of an insufflated abdominal wall. Some camera designs are mounted on abdominal walls by suturing/piercing/magnetic fixation, and controlled by

on-board motors with peripheral mechanisms for 2-DoF rotation [34, 76, 9]. To reduce the design complexity and decrease the camera size, it is preferred to reduce the usages of on-board motors. The cameras' robotic DoFs, such as 2-DoF navigation, 1-DoF rotation (pan motion), can be activated by the magnetic coupling between external magnetic handle and internal magnets [56, 69, 68]. To control the rest 1-DoF for tilt motion in those camera designs, on-board motors are still required.

Efforts have been made on eliminating on-board motors to simplify the actuation mechanism and to reduce power consumption, especially for a wireless laparoscopic camera. Manual manipulation between external magnetic handle and internal magnet can actuate 1-DoF rotation [67]. Automatic control of 2-DoF rotational motion for an insertable camera can be achieved by using electromagnetic coils to drive internal magnet array [40] or using one motorized external magnet to drive orthogonally arranged internal magnets [26]. Both of the two designs work well for close proximity, such as 20 mm abdominal wall thickness. Considering the normal range of abdominal wall thickness, i.e. 20 mm $\sim$ 40 mm [70], the control stability and reliability will be significantly reduced as an abdominal wall thickness increases.

Insertable laparoscopic camera system potentially improves the MIS in multiple aspects: 1) In traditional MIS, it reduces an incision which is specified for insertion of long-stick laparoscopic camera; 2) In Single incision laparoscopic surgeries (SILS), it reduces the port complexity and enhances dexterity of other surgical instruments consequently; 3) Mobile camera systems offers proper and wider view angle of surgical site without imposing limitations for other instruments; 4) Insertable camera system provides proper foundation for implementation of additional systems to improve surgical imaging features such as enhanced surgical illumination [39] and *in-vivo* lens cleaning mechanism [91].

In our prior work [41], we designed a reliable magnetic actuation mechanism for insertable laparoscopic cameras, which enables 2 rotational DoFs under normal range of abdominal wall thickness. The design consists of an external hybrid magnetic driving unit and an internal driven unit. The hybrid magnetic driving unit adopts three permanent magnets and two electromagnetic coils, which are orthogonally arranged as shown in Fig. 2.2. Rotational magnetic field, which actuates the 2-DoF rotational motion of the insertable camera with

the internal driven unit on-board, is generated by spinning the external driving unit and controlling coil input currents. However, in order to accurately actuate the insertable camera to reach its desired orientations, the automatic control of rotation magnetic field generated from the external driving unit is still a major challenge.

In this chapter, we develop an automatic rotational control system for a insertable laparoscopic camera capsule with less than 1° control accuracy, as conceptually illustrated in Fig. 2.1. The contributions of this chapter beyond our prior work include: 1) an abdominal wall thickness estimation system and a current control system were developed for generating appropriate rotational magnetic field to control tilt motion of the camera capsule; 2) an actuation mechanism for the external control unit was designed to enable automatic pan motion control of the capsule; and 3) a fabricated camera system was demonstrated, and experimental analysis was conducted for evaluating the control accuracy of the camera system.

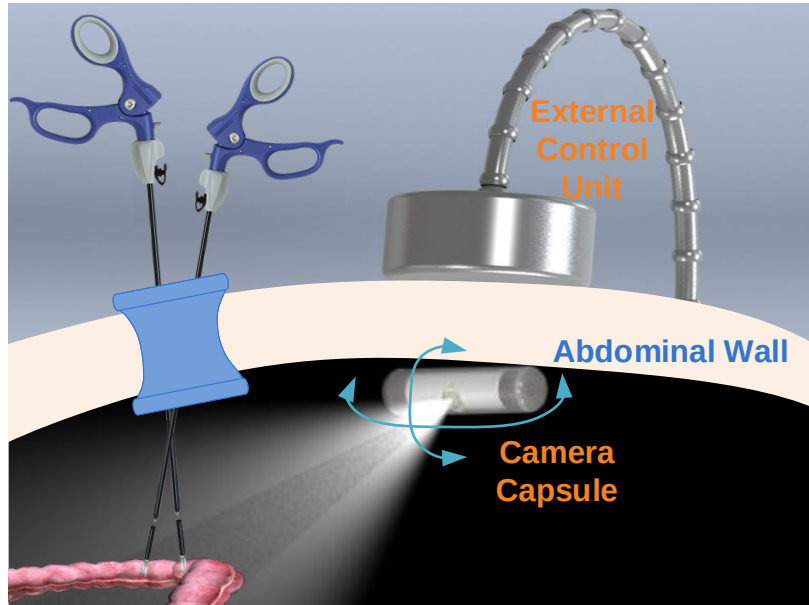


Figure 2.1: Conceptual illustration of the usage of our proposed camera system. The external control unit generate magnetic forces and torques to control the *in-vivo* camera capsule.

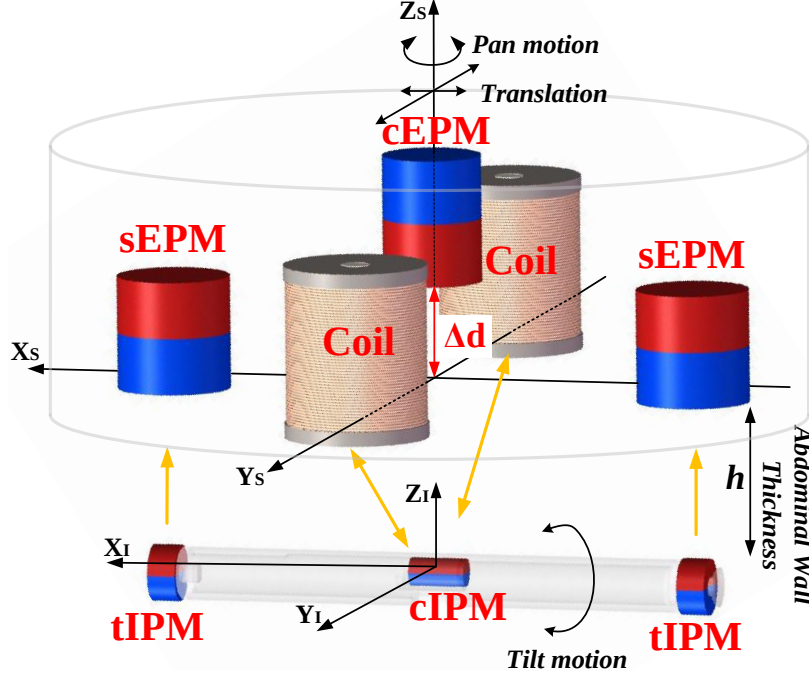


Figure 2.2: Working principle of the camera system.

This chapter is an extension of our preliminary work [38]. Besides providing thorough experimental investigations for the camera capsule's rotation control with statistical analysis, this chapter presents the tilt motion closed-loop control strategy for the capsule, demonstrates a magnetic field calibration system for the external control unit, and provides the calibrated parameters and experimental evaluations for the modeling accuracy. In addition, the control accuracy of the cEPM actuation mechanism is experimentally tested with statistical results. The orientation control system hardware architecture is presented in details. The robustness of the abdominal wall thickness sensing system is also validated by statistical methods.

2.3 Camera Capsule Orientation Control Challenges

The research objective of this chapter is to enable automatic fine orientation control of the camera capsule system, which consists of tilt motion control and pan motion control. The magnetic actuation mechanism designed in our prior work [41] consists of an external driving unit namely a "control unit" and an internal driven unit namely a "camera capsule"

or simply "capsule", as shown in Fig. 2.2. The control unit positioned externally consists of two coils and three permanent magnets, which are orthogonally arranged. The camera capsule positioned internally consists of three diametrically magnetized permanent magnets. The capsule housing that hosts one magnet in the middle can rotate freely related to the other ring magnets at both ends. The pan motion control requires torque along the capsule Z_I axis, while the tilt motion control requires torque along X_I axis.

A spinning motion of the control unit along Z_I can actuate pan motion of the capsule by coupling the magnetic field of the side External Permanent Magnets (sEPMs) and the tail-end Internal Permanent Magnets (tIPMs). The directions of the tIPMs are always aligned with the directions of the sEPMs. Therefore, the capsule is fixed on the internal surface of abdominal wall. The tilt motion can be actuated by the magnetic coupling between the coils and the central Internal Permanent Magnet (cIPM). However, the real situation is that the sEPMs generate strong magnet field at the place where the cIPM locates. Under this condition, the magnetic field from the coils is not strong enough to manipulate the cIPM. The solution is to add a central External Permanent Magnet (cEPM) with the direction opposite to the sEPMs at the control unit center. By adjusting the cEPM displacement along Z_S , the magnetic field from the cEPM can balance out the magnetic field from the sEPMs at the location of the cIPM. The cEPM displacement Δd is adjusted according to the control unit-to-capsule distance for minimizing the magnetic field from the EPMs at the location of the cIPM.

2.4 Control Method of Magnetic Actuation Mechanism

The capsule-to-control unit distance is determined by abdominal wall thickness h . It is required by our system to sense abdominal wall thickness at different locations in real time. Therefore, the **first challenge** to control the camera orientation is how to develop an abdominal wall thickness estimation method and a function $\Delta d = f(h)$. By giving a sensed h , an optimal cEPM displacement $\Delta d = f(h)$ can be calculated with $\Delta d = f(h)$. Based

on this step, the **second challenge** of this chapter is how to develop a closed-loop control scheme for the capsule tilt motion by controlling the coil currents. To generate pan motion of the capsule, the control unit was manually rotated in our previous prototype [41]. In this work, the **third challenge** of this chapter is how to design an automatic pan motion mechanism in the control unit to control the pan motion of camera capsule.

The orientation control architecture of the camera system is illustrated in Fig. 2.3. The control unit magnetic field is initialized by estimating an abdominal wall thickness h (in 2.4.1), calculating an optimal cEPM displacement Δd according to h (in 2.4.2), and adjusting the cEPM to the displacement Δd (in 2.4.3). The input parameters of the orientation control system are the desired tilt angles θ_d and pan angles ϕ_d . The control of tilt motion, which is presented in 2.4.4, involves designing the coil current inputs I_{c1} and I_{c2} to minimize the errors between θ_d and real tilt angle θ . For pan motion, the output angle ϕ is controlled by an actuation mechanism embedded in the control unit, which is presented in 2.4.5.

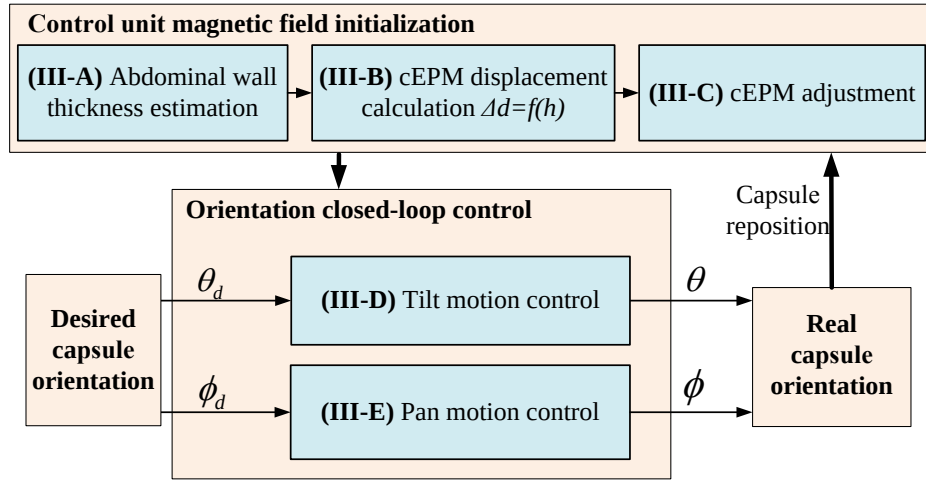


Figure 2.3: The camera system orientation control architecture.

2.4.1 Abdominal Wall Thickness h Estimation

The capsule and the control unit both contact an insufflated abdominal wall internally and externally. The external control unit is anchored and manipulated with a manual goose-neck stand or a robotic arm which provides proper position and orientation of the unit without imposing significant contact force with abdomen. A static capsule with a specific

orientation generates static magnetic field. Based on this fact, the capsule magnetic field measured at the control unit side changes by an abdominal wall thickness. The main idea of estimating an abdominal wall thickness is to sense the capsule magnetic field by using a pre-built capsule magnetic field map, which is with respect to abdominal wall thicknesses or the capsule-to-control unit distances.

There are three challenges to estimate an abdominal wall thickness: 1) which capsule pose can be most effectively used for building a static magnetic field map? 2) how the magnetic field sensors in the control unit can be configured to detect the magnetic field from the capsule? 3) how the sensed data can be used to estimate the capsule-to-control unit distance? The solutions are presented as follows.

Static capsule magnetic field generation

The capsule pose determines the static magnetic field distribution. Fig. 2.4 shows two capsule poses for generating symmetric magnetic field maps. The two capsule poses can be actuated by deactivating the control unit coils and adjusting Δd to its maximum and minimum. With the maximum cEPM displacement shown in Fig. 2.4(a), the cIPM is dominated by the magnetic field from the sEPMs, and aligned with the sEPMs orientation. With the minimum cEPM displacement shown in Fig. 2.4(b), the cIPM is dominated by the magnetic field from the cEPM, and aligned with the cEPM orientation. The arrows in Fig. 2.4(a) and (b) illustrates the magnetic field strength (proportional to the arrow lengths) and magnetic field direction (the arrow directions). It is obvious that the capsule pose in Fig. 2.4(a) can generate more recognizable magnetic field than that in Fig. 2.4(b). Therefore, the control unit initially actuates the cEPM to reach $\Delta d = d_{max}$ for generating the static capsule magnetic field.

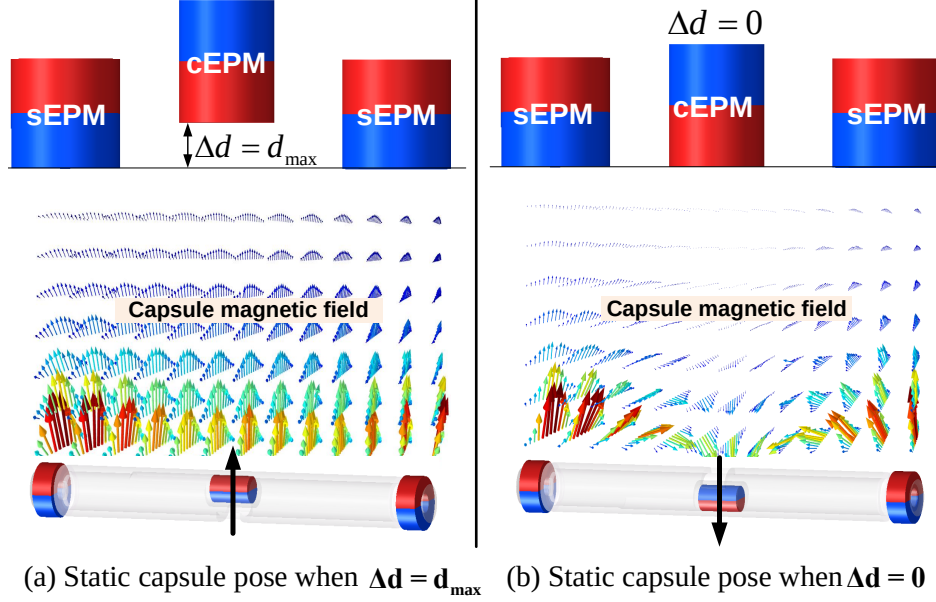


Figure 2.4: Capsule static poses and their magnetic field distributions. (a) Capsule pose 1 with the camera opening pointing up; (b) capsule pose 2 with the camera opening pointing down.

Capsule magnetic field measurement

To measure the magnetic field from the capsule, a set of tri-axis hall effect sensors are installed at the control unit bottom. Because a sensor has its specific measuring range, it is important to select appropriate locations for the sensors to prevent being inundated by the magnetic field from the EPMs. Fig. 2.5(a) shows the magnetic flux density norm distribution of the EPMs at the control unit bottom. The inundation region in dark red is determined by the measuring ranges of the candidate hall effect sensors. In our design, we apply four tri-axis hall effect sensors, which are symmetrically distributed on the control unit bottom. The black squares S_1, S_2, S_3, S_4 represent the sensor locations which are outside the inundation region and closest to the capsule.

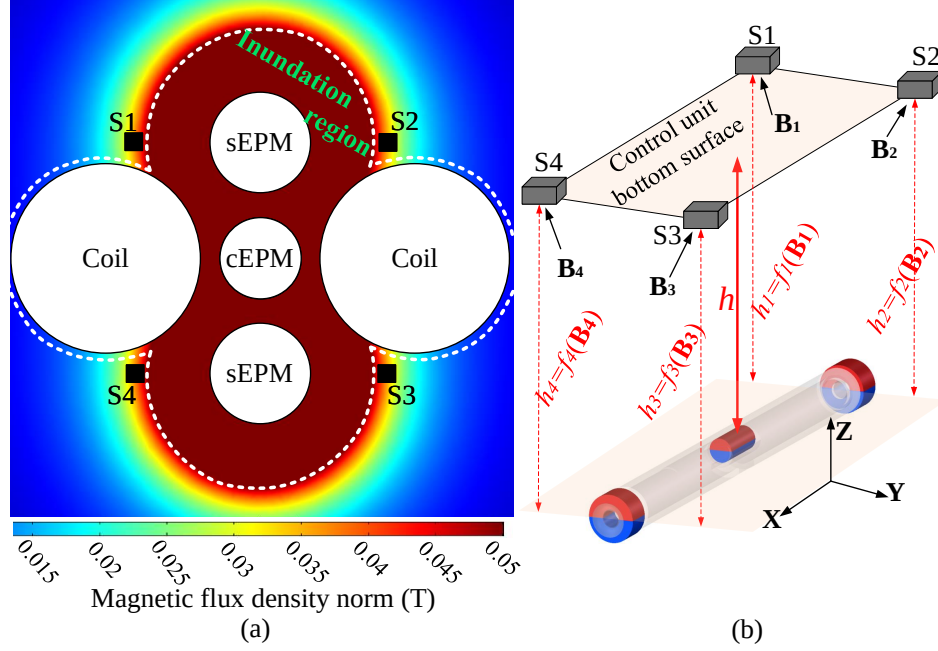


Figure 2.5: Abdominal wall thickness estimation. (a) Magnetic flux density norm of the control unit bottom with the selected sensor locations S_1, S_2, S_3, S_4 ; (b) configurations for developing magnetic field maps which is used to estimate an abdominal wall thickness.

h estimation

Fig. 2.5(b) shows that the tri-axis hall effect sensors detect magnetic field from the capsule at a distance h . Due to the magnetic coupling of the EPMS and the IPMS, the positions between the sensors and the capsule are unchanged in X and Y directions. A magnetic field map f_i represents the relationship between the capsule magnetic field and the capsule-to-control unit distances h_i . The mapping functions f_i represented in lookup tables are developed by recording the magnetic fields with the sensors, while adjusting the capsule-to-control unit distance h .

With the functions f_i and the magnetic fields $\mathbf{B}_i$ sensed by S_i , an abdominal wall thickness can be estimated by

$$h = \frac{1}{N} \sum_{i=1}^N f_i(\mathbf{B}_i), \quad (2.1)$$

where $N = 4$ represents the total group number of the hall effect sensors.

2.4.2 Optimal cEPM Displacement $\Delta d = f(h)$

The cEPM displacement Δd is used to balance out the magnetic field from the sEPMs at the cIPM location. This displacement is effected by an abdominal wall thickness h . Therefore, there is a need to represent Δd as a function of h .

In our prior work [41], the magnetic field from the coils and the EPMs can be analytically modeled by multi-pair magnetic dipoles as follows

$$\mathbf{B} = \frac{\mu_0}{4\pi} m_{00} \Gamma_{00} + \frac{\mu_0}{4\pi} \sum_{i=1}^{K_m} \sum_{j=1}^{N_m} m_{ij} \Gamma_{ij}, \quad (2.2)$$

$$\Gamma_{ij} = \frac{\mathbf{Q}_{ij+}}{|\mathbf{Q}_{ij+}|^3} - \frac{\mathbf{Q}_{ij-}}{|\mathbf{Q}_{ij-}|^3}, \quad (2.3)$$

where m_{00} and m_{ij} represent the strength of magnetic dipoles; $K_m \times N_m + 1$ represents the total number of magnetic dipoles; $\mathbf{Q}_{ij+/-}$ represents a vector from the location of positive/negative magnetic charge to a point in space.

Based on the model in (2.2), the minimized magnetic field $\mathbf{B}_{EPM}$ from the EPMs at the cIPM can be represented by superimposing the magnetic field from each of them.

$$\mathbf{B}_{EPM} = \mathbf{R}_{s1} \mathbf{B}_{s1}(\mathbf{P}_{s1}) + \mathbf{R}_{s2} \mathbf{B}_{s2}(\mathbf{P}_{s2}) + \mathbf{R}_c \mathbf{B}_c(\mathbf{P}_c) = 0 \quad (2.4)$$

where $\mathbf{B}_{s1}$, $\mathbf{B}_{s2}$ and $\mathbf{B}_c$ are the magnetic field models of the sEPMs and the cEPM; $\mathbf{R}_{s1}$, $\mathbf{R}_{s2}$ and $\mathbf{R}_c$ are the rotation matrices to transform the magnetic field into the same coordinate system; $\mathbf{P}_{s1}$, $\mathbf{P}_{s2}$ and $\mathbf{P}_c$ represent the cIPM coordinates in the local frames of the EPMs;

Because h and Δd are both in (2.4), it is desired to explicitly formulate a function as $\Delta d = f(h)$. However, due to the nonlinear property of (2.2) and (2.4), it is difficult to such a function. An alternative way to develop this function is to build a lookup table by giving a range of abdominal wall thickness values h . After searching for Δd that satisfies (2.4), the optimized Δd values are stored in the lookup table. To this end, the optimal vertical displacement of cEPM Δd can be identified in real time for different h values.

2.4.3 cEPM Actuation Mechanism

According to the estimated h from (2.1), the cEPM displacements $\Delta d = f(h)$ are determined in Section 2.4.2. To actuate the cEPM to the displacement Δd , we need to design an adjusting mechanism in the control unit. Considering the strong magnetic coupling between the cEPM and the sEPMs, the mechanism should be able to provide sufficient lifting force for the cEPM, and keep the whole control unit as compact as possible.

Fig. 2.6 illustrates the design of the cEPM actuation mechanism in the control unit. The design has two major considerations: 1) the actuation force for the cEPM movement can be efficiently provided by the worm and gear mechanism; 2) the self-locking function of the design enables the cEPM to keep still when the motor is unactuated. The bolt that connects with the cEPM by using a coupler is actuated by the gear rotation. The bolt track restricts the bolt rotation, and keeps the bolt and the cEPM moving vertically.

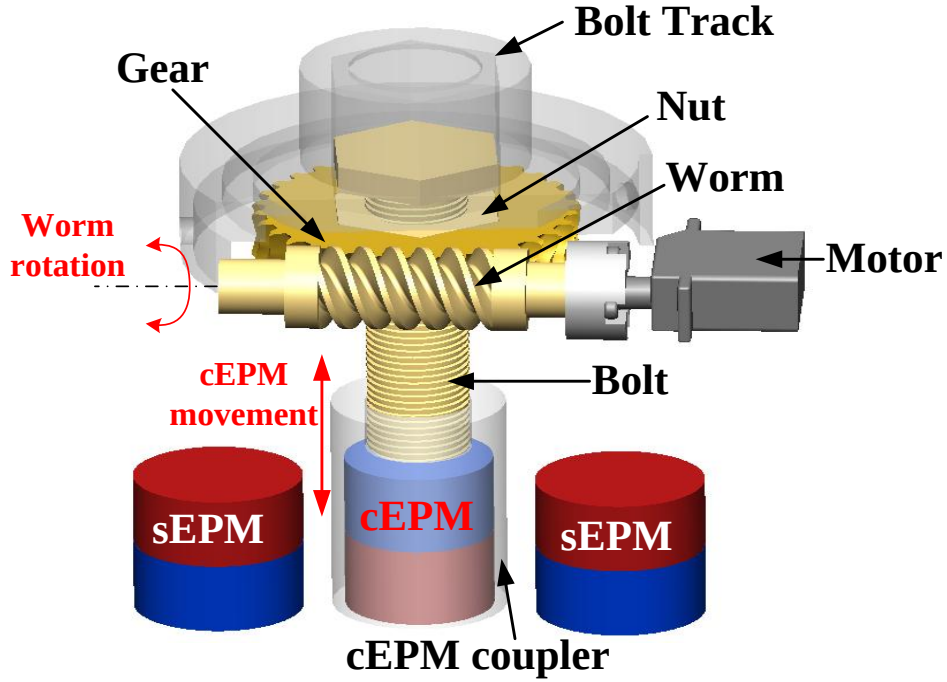


Figure 2.6: cEPM displacement actuation mechanism in the control unit.

2.4.4 Tilt Motion Control

Capsule protection shell

The camera design in our prior work [41] may potentially get blurred by peritoneal fluid due to the direct contact with an abdominal wall. To address this issue, a transparent shell is applied to prevent the camera lens from contacting with tissue and also to maintain imaging quality, as illustrated in Fig. 2.7(a)-5. Two ceramic bearings (Fig. 2.7(a)-7) and two tIPM shafts (Fig. 2.7(a)-6), which are fixed in the shell caps (Fig. 2.7(a)-1), are used for hanging the camera housing (Fig. 2.7(a)-3) inside the transparent shell with no contact. This design provides the camera smooth rotation inside the shell when rotational torque is exerted on the cIPM (Fig. 2.7(a)-4). Another benefit of this design is to make the laparoscopic camera reusable by depositing the shell after use, and subsequently reduces a surgery cost.

The new capsule design insulates the camera lens from contacting with an abdominal wall. It makes the capsule housing free of frictional torque from an abdominal wall during tilt motion. Compared with the tilt motion control method developed in [41], this new capsule design can significantly reduce the motion control complexity and improve the control robustness, which has been experimentally proved in Section 2.5.9. The detailed information to control the tilt motion of the new capsule are as follows.

Control with electromagnetic coils

The camera tilt motion is activated by the magnetic coupling between the coils and the cIPM. I_{c1} and I_{c2} represent current inputs of the coils, and θ represents a tilt angle.

The factors that impact on the camera tilt motion need to be firstly considered. After estimating an abdominal wall thickness h and actuating the cEPM displacement Δd , the EPMS have ignorable impact on the tilt motion of the cIPM. Beside the control inputs I_{c1} and I_{c2} , the tilt motion of the cIPM is also affected by T_g and T_{tIPM} which are the torques due to the robot gravity along X_I and the magnetic torque on the cIPM from the tIPMs along X_I respectively. As illustrated in Fig. 2.7(b), T_g and T_{tIPM} are always in opposite directions, and are canceled by each other according to our preliminary experimental investigation. Therefore, the camera tilt angle aligns with the direction of the magnetic field generated by

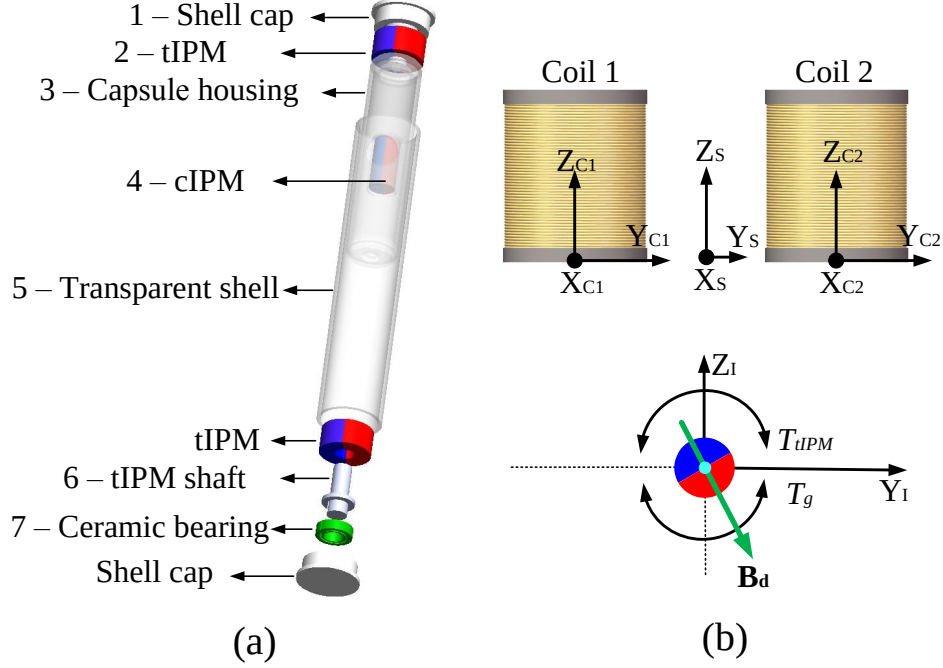


Figure 2.7: (a) Conceptual illustration of the capsule design. (b) Capsule tilt motion control with the coils.

the coils. The objective to control the camera tilt motion is to determine I_{c1} and I_{c2} for generating a magnetic field with the direction θ at $\mathbf{P}_s$ in $\Sigma_S\{X_S, Y_S, Z_S\}$, where $\mathbf{P}_s$ is the location of the cIPM.

The relationship between the current inputs and desired magnetic field directions can be formulated by (2.5) in Σ_S as follows

$$\underbrace{(\mathbf{R}_{C1}\mathbf{B}_c^u)}_{\mathbf{B}_{c1}^S} I_{c1} + \underbrace{(\mathbf{R}_{C2}\mathbf{B}_c^u)}_{\mathbf{B}_{c2}^S} I_{c2} = \underbrace{\mathbf{R}_I^S \mathbf{R}_{Ix}(\theta) \mathbf{B}^{-z}}_{\mathbf{B}_d}, \quad (2.5)$$

where $\mathbf{R}_I^S \in \mathbb{R}^{3 \times 3}$ represents a rotation matrix from $\Sigma_I\{X_I, Y_I, Z_I\}$ to Σ_S . According to the setting of Σ_S and Σ_I in Fig. 2.7(b), $\mathbf{R}_I^S$ is an identity matrix. $\mathbf{R}_{Ix}(\theta) \in \mathbb{R}^{3 \times 3}$ represents the rotation matrix along X_I with θ as a variable. $\mathbf{B}^{-z} \in \mathbb{R}^{3 \times 1}$ denotes a unit vector pointing in $-Z_I$. $\mathbf{B}_{c1}^S, \mathbf{B}_{c2}^S \in \mathbb{R}^{3 \times 1}$ denote the unit-current magnetic field of the coils at $\mathbf{P}_s$ in Σ_S . $\mathbf{B}_d \in \mathbb{R}^{3 \times 1}$ denotes the desired magnetic field direction as shown in Fig. 2.7(b). Benefiting from the magnetic field model in (2.2), $\mathbf{B}_{c1}^S, \mathbf{B}_{c2}^S$ can be computed in real time.

Considering the x components in $\mathbf{B}_{c1}^S$, $\mathbf{B}_{c2}^S$, and $\mathbf{B}_d$ are zeros, (2.5) is reformulated as

$$\mathbf{B}_d^{yz} = \mathbf{B}_c^{S_{yz}} \mathbf{I}_c, \quad (2.6)$$

where $\mathbf{B}_c^{S_{yz}} = [\mathbf{B}_{c1}^{S_{yz}}, \mathbf{B}_{c2}^{S_{yz}}] \in \mathbb{R}^{2 \times 2}$, $\mathbf{B}_d^{yz} \in \mathbb{R}^{2 \times 1}$, $\mathbf{I}_c = [I_{c1}, I_{c2}]^T \in \mathbb{R}^{2 \times 1}$. Because $\mathbf{B}_c^{S_{yz}}$ has a full row rank, the current input vector $\mathbf{I}_c$ can be found in (2.7) by applying pseudoinverse to (2.6)

$$\mathbf{I}_c = (\mathbf{B}_c^{S_{yz}})^T (\mathbf{B}_c^{S_{yz}} (\mathbf{B}_c^{S_{yz}})^T)^{-1} \mathbf{B}_d^{yz}. \quad (2.7)$$

To achieve high accuracy tilt motion control, a closed-loop control scheme is developed as illustrated in Fig. 2.8. Given a desired tilt angle θ_d , a current vector is calculated by $\mathbf{I}_c = \tilde{f}(\mathbf{P}_s, \tilde{\theta})$ which is another representation of (2.7). Because the solution of (2.7) only provides the ratio of I_{c1} and I_{c2} , the input current vector is amplified with I_{max} to maximize the generated magnetic field. A current input trajectory $\mathbf{I}_c^{max}(t)$ can thus be developed by a given time period ΔT , and applied to the capsule. The arrived tilt angle θ at the end of $\mathbf{I}_c^{max}(t)$ is measured by a tilt angle sensor for feedback. If $\delta = |\theta_d - \theta|$ is smaller than a given threshold δ_θ , the tilt angle control process is complete. Otherwise, an adjusted tilt angle $\tilde{\theta} = \theta_d \pm \Delta\theta$, where $\Delta\theta$ is a small step angle, is served as a new input to calculate $\mathbf{I}_c$.

2.4.5 Pan Motion Mechanism

To control the pan motion automatically, it is desired to design an actuation mechanism that can generate rotational motion of the control unit. Fig. 2.9 conceptually demonstrates the pan motion mechanism in the control unit. The mechanism controls rotational motion of the control unit core, which is sitting inside the control unit shell. A motor driven spur gear

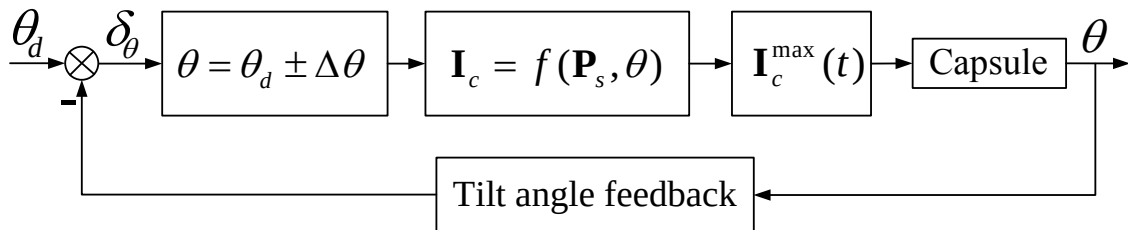


Figure 2.8: Block diagram of the camera tilt motion control.

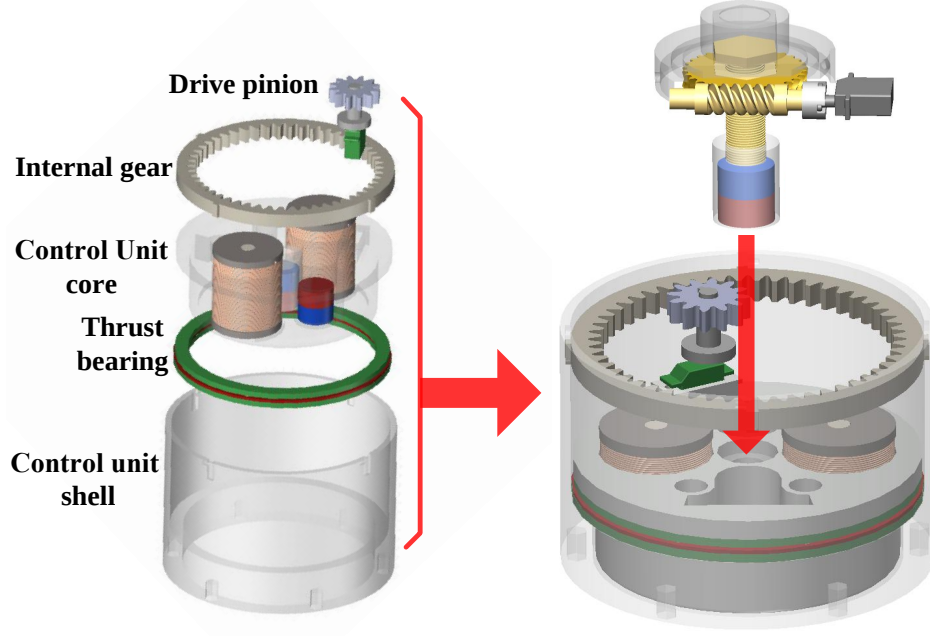


Figure 2.9: Pan motion actuation mechanism in the control unit.

is fixed on the control unit core, while an internal gear is attached on the internal surface of the control unit shell. The control unit core can thus be actuated by the relative motion of the spur gear and the internal gear. To keep smooth relative rotation between the control unit core and the control unit shell, a thrust bearing is applied at their contact surfaces. The cEPM actuation mechanism presented in 2.4.3 is fixed on the top of the control unit core, and the cEPM is inserted in the middle hole on the control unit core.

2.5 Prototype Development and Experimental Investigation

2.5.1 Experimental Platform

Fig. 2.10(a) shows the overview of the experimental environment. The major parts of the camera system were fabricated by a 3D prototyping machine (Fortus 400mc, Stratasys Inc.). To simulate the viscoelastic properties of a real insufflated abdominal wall (average Young's modulus 32.5 kPa) [70], a viscoelastic material Durometer 40 (Young's modulus 27.57 kPa at 15% deflection, Sorbothane, Inc.) was applied. The initial abdominal wall thickness was

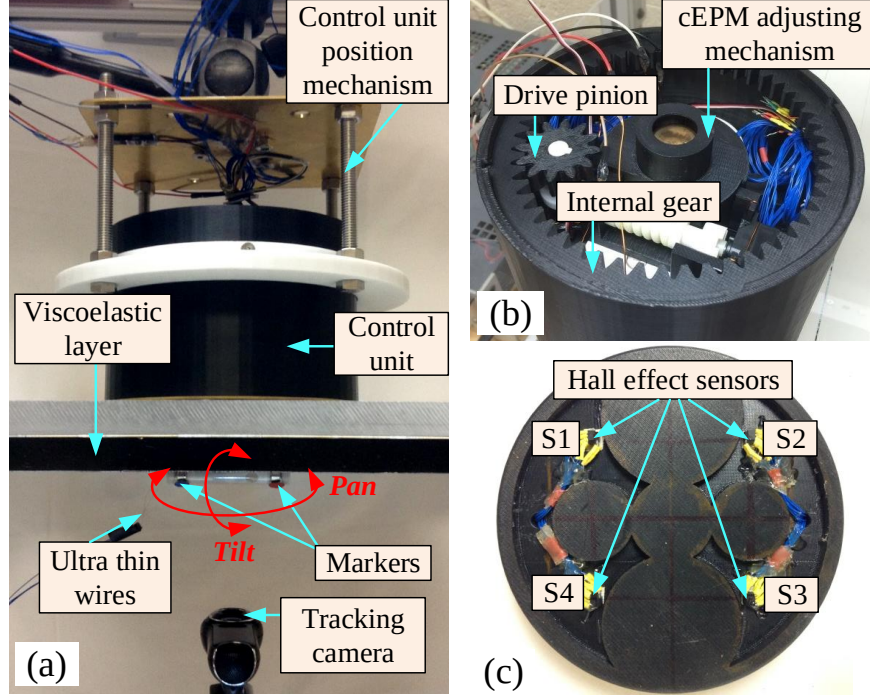


Figure 2.10: Experiment platform setup and the control unit configuration. (a) Overview of the camera orientation control system. (b) Demonstration of the pan motion mechanism and the cEPM actuation mechanism. (c) Magnetic field sensing system.

26 mm (tissue layer 15 mm, support layer 11 mm). To increase the abdominal wall thickness for experimental studies, a vertical control unit positioning mechanism in Fig. 2.10(a) was applied. A silicone oil lubricated capsule-tissue contact layer was added to the bottom of the viscoelastic material for mimicking an internal abdominal wall surface.

Control system hardware architecture

Figure 2.11 shows the hardware architecture for the camera orientation control system. A tethered current control system was developed based on Pulse Width Modulation (PWM). The system consists of a micro-controller (STM32F4Discovery, STMicroelectronics Inc.) to generate PWM signals, two PWM amplifiers (L6205 DMOS Full Bridge Driver, STMicroelectronics Inc.) to amplify the signals, a power supply for powering up the amplifiers, and a PC to send control command to the micro-controller via serial communication. The abdominal wall thickness sensing system was implemented by setting four sets of tri-axis hall effect sensors to detect magnetic field from the capsule. The detected analog magnetic

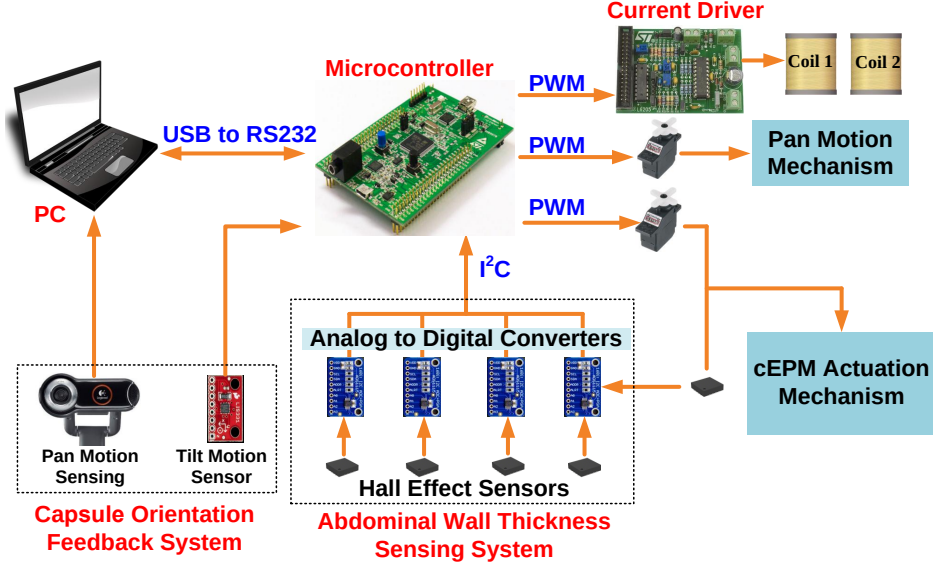


Figure 2.11: Control system hardware architecture.

field signals were converted into digital signals through four sets of AD converters. In order to provide closed-loop feedback sensing, at this stage we applied an external camera and on-board sensor to acquire real time orientation feedback.

2.5.2 Control unit fabrication

cEPM adjusting mechanism

The fabrication of the cEPM adjusting mechanism that proposed in section 2.4.3 involves two issues: 1) the mechanism has to provide sufficient lifting force to overcome the magnetic force between the cEPM and the sEPMs; 2) the cEPM has to be accurately controlled to the desired displacement Δd .

The design objective is to make the cEPM adjusting mechanism compact enough to fit in the control unit shell (external diameter 164 mm and height 109 mm). The worm in Fig. 2.12(a) is with pitch diameter 12.19 mm, outside diameter 15 mm, number of threads $N = 3$, worm length 25 mm. The gear is with 32 teeth, pitch diameter 45 mm, outer diameter 49.25 mm, face width 10 mm. The nut that fixed at the gear center is used for generating vertical displacements of the bolt by rotating the gear and the nut. The hex nut is with height 13.89 mm, diameter 15.875 mm. The bolt is matching with the nut with thread length

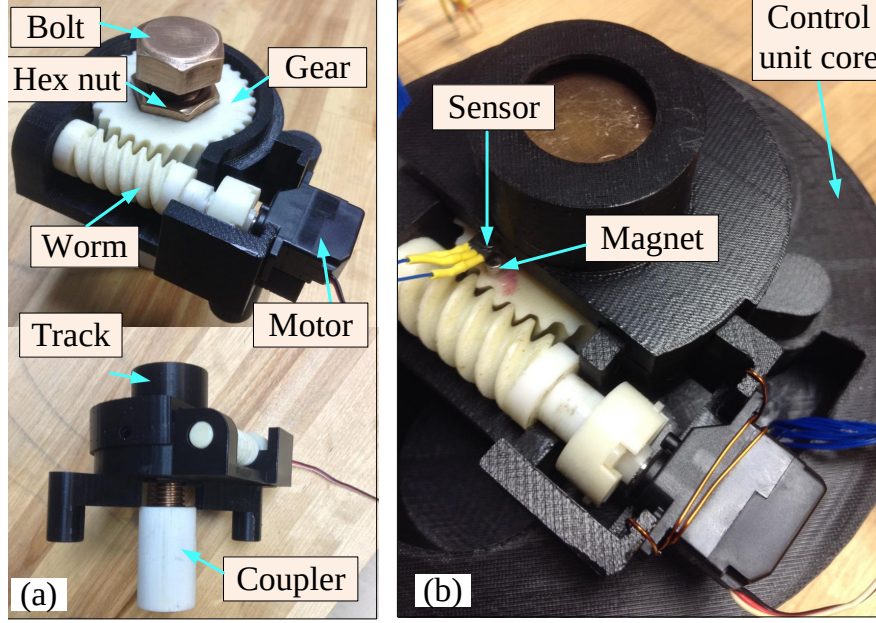


Figure 2.12: Configurations of the cEPM adjusting mechanism.

53 mm, screw pitch $p = 2$ mm. According to our preliminary experiments, the maximum force needed to pull out the cEPM from the magnetic coupling between the cEPM and the sEPMs is about $F = 50$ N. The required motor torque M_T can be estimated by

$$M_T = \frac{F \cdot p}{2\pi \cdot \eta_s \cdot \eta_{wg} \cdot r}, \quad (2.8)$$

where η_s and η_{wg} are the efficiencies of the screw and the worm-gear respectively; $r = 32/3$ is the ratio of the worm gear. M_T is estimated as 0.149 Nm by conservatively assuming $\eta_s = 10\%$ and $\eta_{wg} = 10\%$. A servo motor (S3156, Futaba Inc.) in Fig. 2.12(a), which can provide 0.196 Nm at 4.8 V, was modified into continuous rotation for our application. To avoid influencing the magnetic field of the control unit, the silicone brass screw and nut (Bolt Depot, Inc.) were selected for the design.

Fig. 2.12(b) shows the assembly of the cEPM adjusting mechanism with the displacement sensing system on the control unit core that houses the EPMS and the coils. To sense the cEPM vertical displacements, a magnetic encoder is designed by using four tiny cylindrical permanent magnets symmetrically distributed on the gear surface. A single hall effect sensor (CYL8405, Chenyang-Technologies GmbH& Co. KG) is applied to pickup the magnetic field

signal from the magnets to measure the gear rotation. The screw travels 2 mm when the gear generates a full rotation. Four magnets can thus provide 0.5 mm control resolution.

Abdominal wall thickness sensing system

According to the design in Section 2.4.1, the magnetic field sensing system were implemented at the bottom of the control unit core, as illustrated in Fig. 2.10(c). The bottom of the control unit core is carved around the EPMS and the coils with 5 mm depth for installing magnetic field sensors. Four sets of tri-axis hall effect sensors with measuring ranges $0 \sim 64$ mT and sensitivities about 50 mV/mT were symmetrically fixed in the carved bottom. To enable the system to measure tiny changes of the capsule magnetic field, the measured signals were converted to digital signals with sensing resolution 0.01 mT by using a 16-bit ADC (ADS1115, Texas Instruments Inc.).

The estimation function in (2.1) was implemented by the data acquired from the experimental setup in Fig. 2.13(a), which has the same sensor configuration as that in the control unit in Fig. 2.10(c). The capsule with the pose demonstrated in Fig. 2.4(b) was fixed on a vertical positioning stage for adjusting the distance between the capsule and the sensors. Fig. 2.13(b) shows the estimation functions f_i , ($i = 1, 2, 3, 4$), which record the norms of the magnetic flux densities $|\mathbf{B}_i|$ and the vertical distances h_i between the capsule and the sensors S_i . An average value of h_i was used as the final estimated abdominal wall thickness h to improve the robustness of this method.

2.5.3 Control unit dimensions

The fabricated control unit is demonstrated in Fig. 2.10(b). The internal gear has 52 teeth with pitch diameter 132 mm, major diameter 139.4 mm, and minor diameter 128 mm. The drive pinion has 12 teeth with pitch diameter 30.48 mm, major diameter 36.56 mm, and minor diameter 25.13 mm. The face widths of the internal gear and the drive pinion are both 10 mm. The control unit shell is with external diameter 164 mm and height 109 mm. The thrust bearing with inner diameter 133.6 mm and outer diameter 154.6 mm sits inside the control unit shell with 30 mm distance to the shell bottom. A modified digital servo motor,

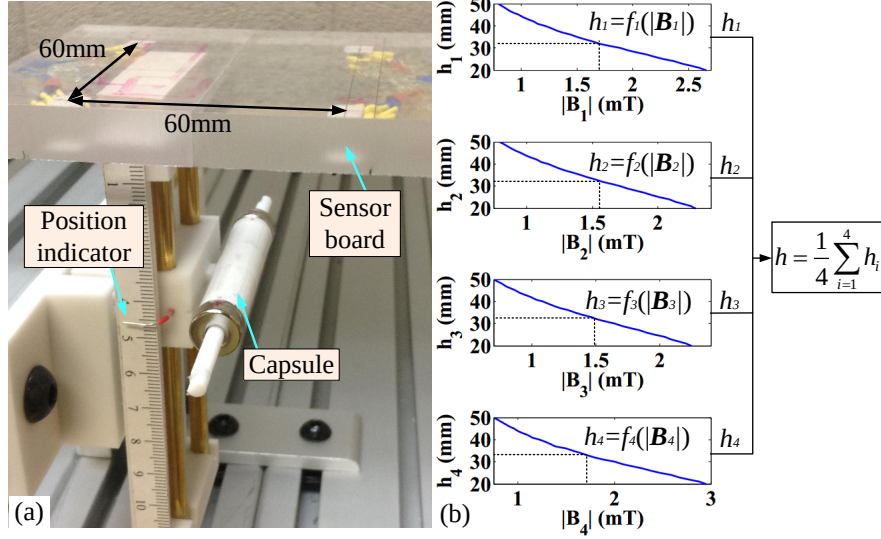


Figure 2.13: Experimental configuration for developing abdominal wall thickness estimation functions.

which is identical to the motor applied on the cEPM adjusting mechanism, is fixed in the control unit core and connected with the drive pinion. A brass board with diameter 164 mm and thickness 1.64 mm was applied at the bottom of the control unit shell for protecting the control unit core and avoiding magnetic field influence between the capsule and the control unit.

2.5.4 Magnets and coil selection

The magnets applied in the control unit are all axially magnetized cylindrical shape with two different dimensions. The sEPMs are 25.4 mm in diameter, 25.4 mm in length, and 1.43 T magnetic remanence. The cEPM is 22.22 mm in diameter, 28.57 mm in length, and 1.43 T magnetic remanence. The magnets in the capsule are all diametrically magnetized. The tIPMs are ring-shaped with 12.7 mm in outer diameter (OD), 4.75 mm in inner diameter (ID), 6.35 mm in thickness, and 1.32 T magnetic remanence. The cIPM is cylindrical shape with 6.35 mm in diameter, 12.7 mm in length, and 1.32 T magnetic remanence (K&J Magnetics, Inc.). Two iron-core coils are applied in our design with OD 50 mm, ID 10 mm, height 50 mm, and 2000 winding turns.

2.5.5 Capsule dimensions

Fig. 2.14 shows the fabricated dummy camera with the disassembled parts. The outer diameter of the transparent shell (MOCAP, Inc.) determines the camera diameter as 14.52 mm which can fit in a standard trocar with diameter 12 mm ~ 15 mm. The miniature ceramic ball bearings (NationSkander California Corp.) enable smooth rotation of the camera housing inside the transparent shell. The bearings are 5 mm in inner diameter, 9 mm in outer diameter, and 3 mm in thickness.

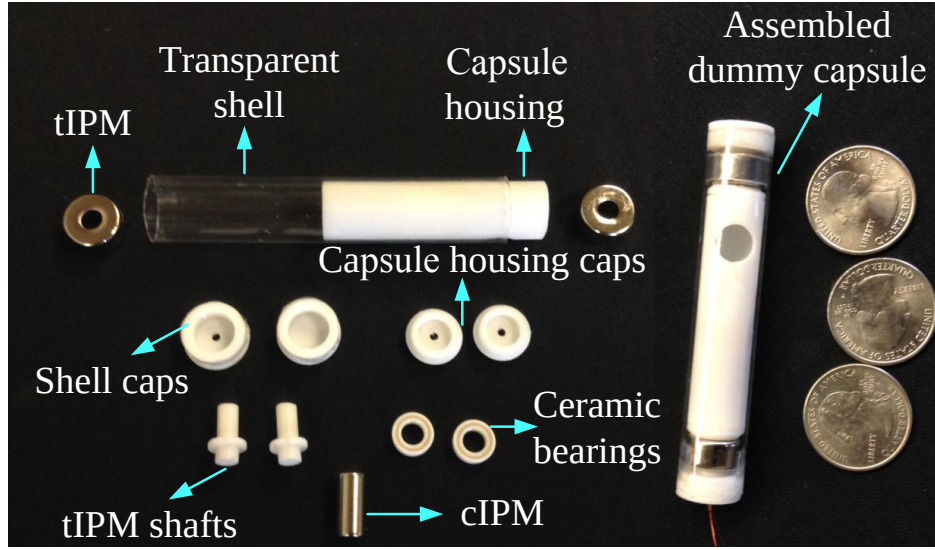


Figure 2.14: The disassembled dummy capsule parts and the assembled dummy capsule.

2.5.6 Calibration of Magnetic Field Models

The control unit magnetic field model developed in Section 2.4.2 were formulated by (2.2) with parameters m_{00} , m_{i1} , ($i = 1, \dots, 10$). Due to the imperfection of the coil wrapping, the iron cores, and the permanent magnets, the model calibration was performed by using experimental data. Fig. 2.15(a) shows the experiment setup for magnetic field calibration, which consists of a transparent board to support the EPMS or the coils, one X-Z position stage, and hall effect sensors for three axes magnetic field sensing. Fig. 2.15(b) and (c) show the magnetic flux density norm comparison results of the experimental data and the magnetic field models.

Fig. 2.15(b) illustrates the calibrated coil magnetic field model (the blue line) with unit-current input based on the experiment data (the blue circles). The red line and the red circles illustrate the magnetic field comparison result between the predicted magnetic field with calibrated parameters in Table 2.1 and the sensed experiment data when $I = 0.5$ A. This result validates the linear relationship between the coil input current and the generated magnetic field strength. Fig. 2.15(c) shows the magnetic field comparison result between the EPMS magnetic field model prediction data and experiment data with the cEPM vertical displacements $\Delta d = 1$ mm, 3 mm, 5 mm, 7 mm. The coincidences of the model prediction data and the experiment data at $B_{norm} = 0$ with various Δd validate the optimal displacement function $\Delta d = f(h)$ developed in Section 2.4.2. The average errors of the magnetic field models demonstrated in Fig.2.15(b) and (c) were 0.11% and 0.23% respectively. The calibrated parameters of the EPMS and the coils are shown in Table 2.1.

Table 2.1: Calibrated parameters of the sEPMS, the cEPM and the coils. Units: $[A \cdot m^2]$

	m_{00}	m_{11}	m_{21}	m_{31}	m_{41}	m_{51}	m_{61}	m_{71}	m_{81}	m_{91}	$m_{10,1}$
sEPM	2.479	2.676	2.516	2.294	2.063	1.894	1.759	1.536	1.225	1.136	1.252
cEPM	2.493	2.456	2.436	2.393	2.349	2.305	2.264	2.233	2.209	2.193	2.185
Coil	2.494	2.401	2.394	2.398	2.394	2.403	2.395	2.398	2.398	2.393	2.401

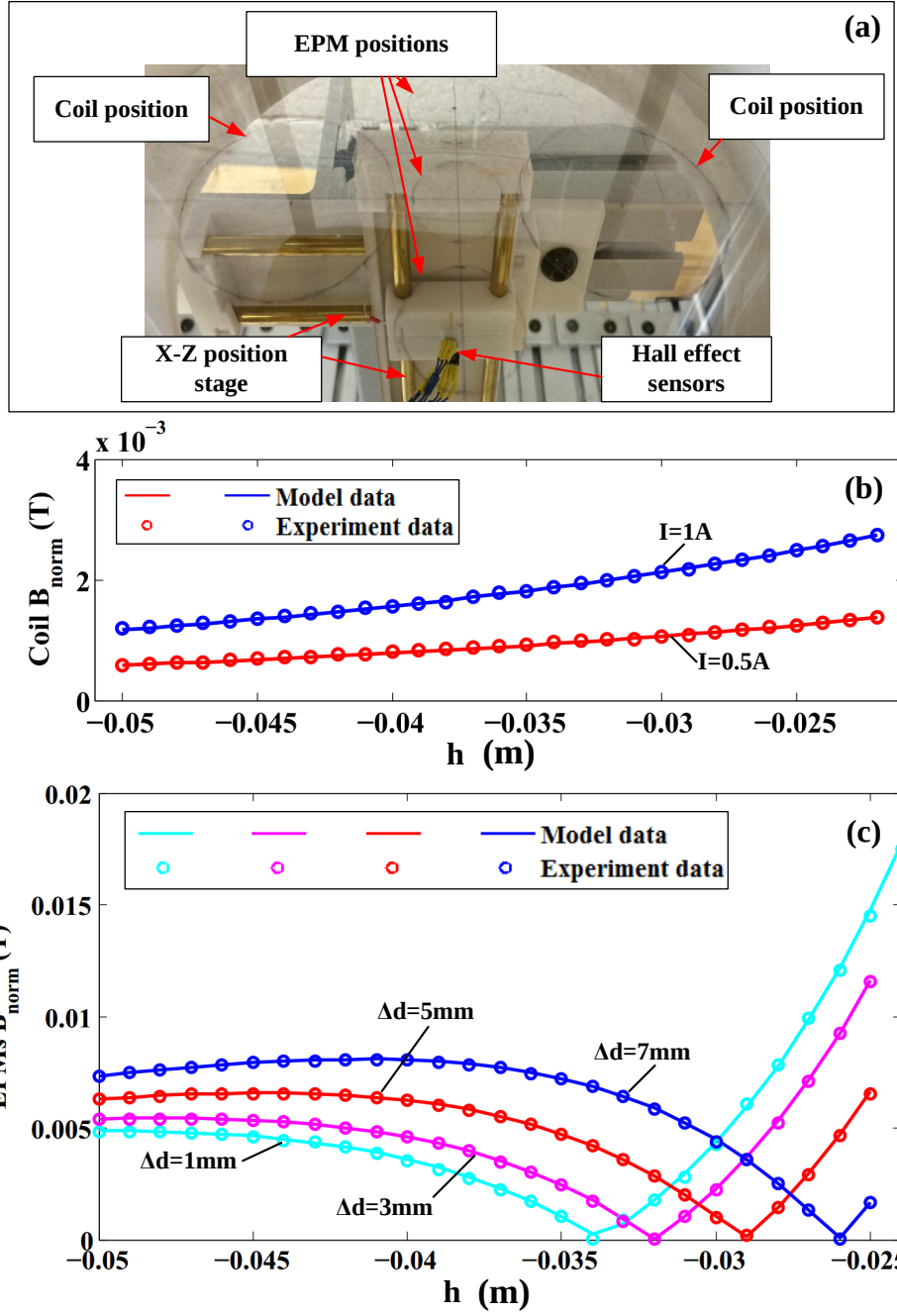


Figure 2.15: Experimental setup for calibrating the magnetic field models of the control unit. (a) Magnetic field calibration platform with a supporting board for magnets/coils, X-Z position stage, and magnetic field sensors. (b) Experimental results for the coils calibration. (c) Experimental results for the EPMs calibration.

2.5.7 Control of cEPM Displacements

Fig. 2.16 shows the investigation of the control accuracy for the cEPM adjusting mechanism developed in Section 2.5.2. The measured displacement from a caliper served as the benchmark. The desired cEPM displacements were set from 2 mm to 10 mm with 2 mm intervals. For each test, the cEPM displacement was initially set at $\Delta d = 0$. The results show that the maximum errors of all group tests were within 0.4 mm, which provides sufficient control accuracy for the camera system. As the cEPM displacements traveled from 2 mm to 10 mm, the errors were accumulated. The accumulated error can be limited within 0.37 mm by resetting the displacement counter when the minimum or the maximum displacement is reached.

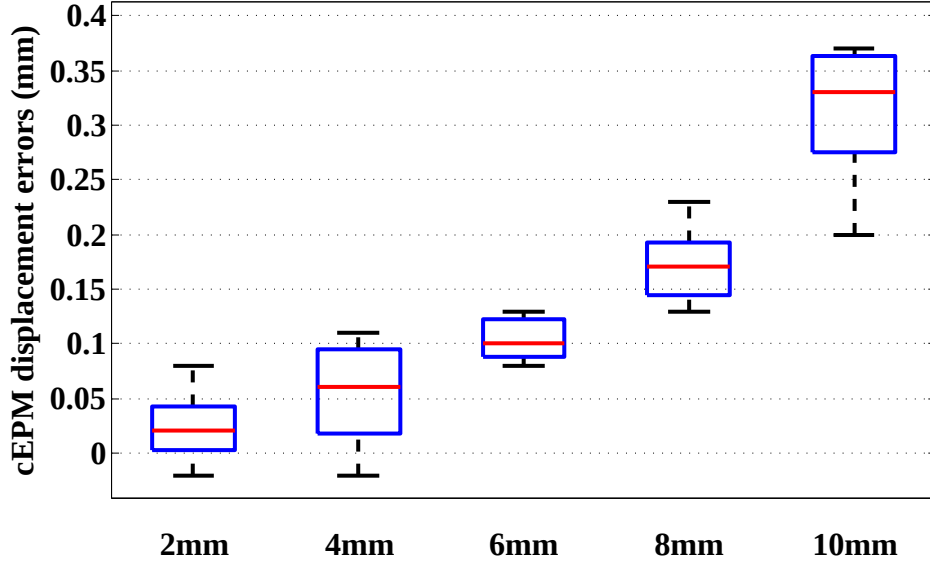


Figure 2.16: Investigation of the cEPM displacement control accuracy. Desired cEPM displacements Δd were set from 2 mm to 10 mm.

2.5.8 Evaluation of Abdominal Wall Thickness Estimation Method

Fig. 2.17 shows the experiment setup and experiment results for evaluating the abdominal wall thickness estimation method developed in Section 2.4.1. To investigate the accuracy and robustness of the proposed method, the experiment was divided into two groups, which are 1) the estimation under the impact of the EPMS; and 2) the estimation without the impact from the EPMS. The two-group experimental setups can be achieved by putting or removing

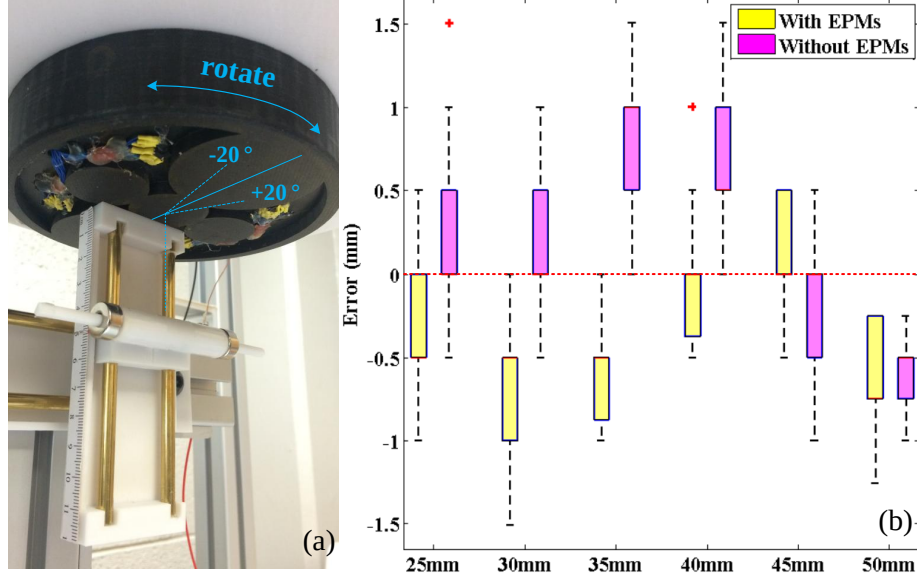


Figure 2.17: Abdominal wall thickness estimation method investigation. (a) Experimental setup. (b) Two-group experimental results for the estimations with/without the magnetic field influence from the EPMs.

the EPMs in the control unit core with the coils deactivated, as shown in Fig. 2.17(a). In both experiment groups, the capsule with the pose demonstrated in Fig. 2.4(a) was attached to a Z-axis positioning stage under the control unit core. To test the estimation robustness, the control unit core was rotated within the range of $\pm 20^\circ$ during each estimation process. Fig. 2.17(b) shows the capsule-to-control unit distance estimation errors at 25 mm, 30 mm, 35 mm, 40 mm, 45 mm, and 50 mm. The mean absolute errors (MAEs) and the standard deviations (SDs) of the two-group experiments under different testing distances are shown in Table 2.2. The data show the abdominal wall thickness estimation system can provide sub-millimeter accuracy. To investigate the differences of the estimation errors between the two-group experiments, two-sample t-tests were applied to compare the MAEs and the SDs. The P values for the MAEs and the SDs were 0.6693 and 0.4210 respectively, which indicate that the differences were considered to be statistically insignificant.

2.5.9 Camera Orientation Control

The closed-loop control of the camera motion requires sensing systems for tilt angles and pan angles. Due to the lack of on-board wireless internal sensor at the current stage,

Table 2.2: Mean Absolute Error (MAE) and Standard Deviation (SD) of The Estimated Capsule-to-control unit Distances

Dist. (mm)	With EPMs		Without EPMs	
	MAE (mm)	SD (mm)	MAE (mm)	SD (mm)
25	0.2672	0.3531	0.2368	0.5506
30	0.6118	0.3025	0.1771	0.2989
35	0.5256	0.3431	0.7216	0.3678
40	0.0116	0.3856	0.6910	0.3807
45	0.2	0.2857	0.1684	0.3832
50	0.5682	0.2424	0.5729	0.1943

separate tethered motion feedback systems were designed for pan and tilt motion, as shown in Fig. 2.11. The tilt motion sensing system applied a tri-axis accelerometer (LIS331HH, STMicroelectronics Inc.), which is fixed inside the camera housing. To minimize the impact on the camera tilt motion from the wires, 42 AWG enameled copper wires with diameter 0.066 mm (Remington Industries, Inc.) were applied for power supply and data transmission. The pan motion sensing system applied a webcam (Logitech Pro 9000) under the capsule to track the positions of two color markers on the transparent shell of the dummy camera. The current inputs of the two coils are limited at $|I_{max}| = 1.5$ A to prevent coil overheating.

The camera orientation control resolution depends on the error thresholds for tilt motion (δ_θ) and pan motion (δ_ϕ). An excessively fine orientation control resolution, such as 0.1° , is not only unnecessary for a laparoscopic surgery, but it also challenges the system real time performance. In this work, δ_θ and δ_ϕ are both set as 1° to achieve a balance between high control resolution and burden on system real time performance.

Fig. 2.18 and Fig. 2.19 demonstrate the control accuracies of the orientation control system under two different abdominal wall thicknesses, which are 26 mm and 40 mm. To

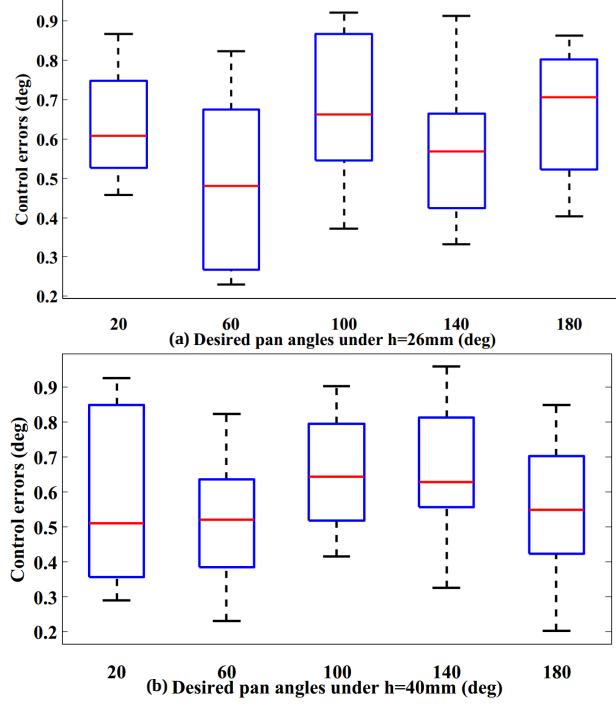


Figure 2.18: Pan motion closed-loop control by setting desired angles from 20° to 180°. (a) under 26 mm abdominal wall thickness; (b) under 40 mm abdominal wall thickness.

test the pan motion control accuracy, desired angles are set at 20°, 60°, 100°, and 120° with initial angles set at 0°. 10 trials were executed for each desired control angle. The statistic results in Fig. 2.18 show that the camera pan motion can be successfully controlled with an average control error 0.594°. To test the tilt motion control accuracy, desired angles are sampled from 10° to 70° with an interval as 10°.

Fig. 2.19 shows that all the control errors of tilt motion are all limited within 1° with an average error as 0.524°. Fig. 2.20 shows that the tilt motion control trajectories for the desired tilt angles at 15°, 45°, and 75° which took 0.93s, 1.74s, and 1.99s to reach steady states.

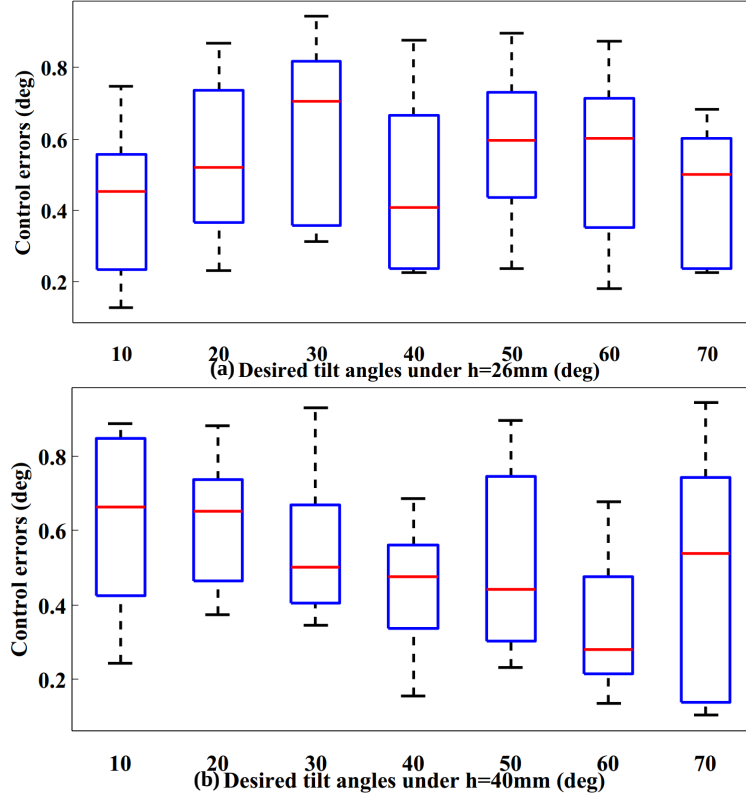


Figure 2.19: Tilt motion closed-loop control by setting desired angles from 20° to 180°. (a) under 20 mm abdominal wall thickness; (b) under 40 mm abdominal wall thickness.

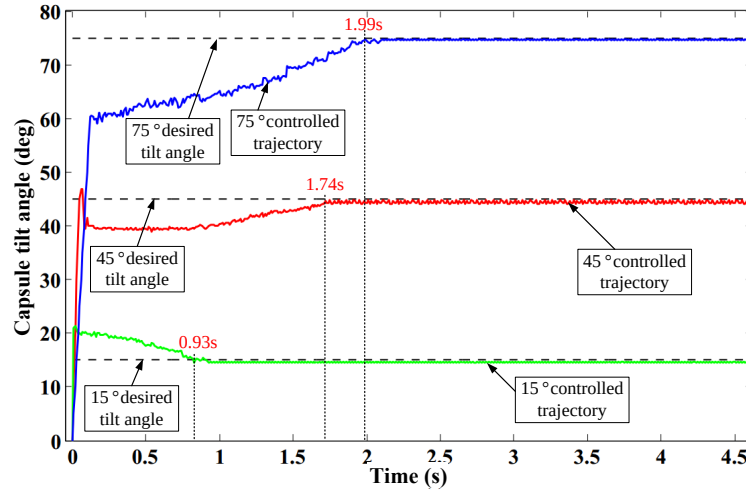


Figure 2.20: Tilt motion control trajectories under an abdominal wall thickness as 40 mm.

2.6 Discussion

As shown in Fig. 2.15(c), the optimal displacement of the cEPM is $\Delta d = 1$ mm when the abdominal wall thickness is about 34 mm. Because the minimum value of Δd is zero, the cEPM cannot be further adjusted when the abdominal wall thickness is over 35 mm. However, this does not result in a problem to control the camera tilt motion. The cEPM displacement $\Delta d = 0$ and the increased abdominal thickness work together to reduce the magnetic field on the cIPM from the EPMs. The effectiveness of the tilt motion control with an 40 mms abdominal wall thickness has been validated in Fig. 2.19(b). Combined tilt and pan actuation of capsule was experimented for 0° to 75° tilt motion and 0° to 180° pan motion, results are shown in Fig. 2.21.

To enable sufficient range of motion for inspecting an abdominal cavity, a minimum range of the camera tilt motion is required. This range of motion depends on the field of view of an on-board camera module that will be integrated in our future work. For example, by conservatively assuming the camera field of view as 50° , the minimum required tilt angle is 65° , which can be easily achieved by our design as illustrated in Fig. 2.19.

Considering some patients, who are obese or skinny, with their abdominal thicknesses out of the normal range 20 mm $\sim$ 40 mm, the prototype design in this chapter may not function appropriately. In that case, the control unit parameters can be designed according to the largest abdominal wall thickness, such as 70 mm. To make the robot system function normally when an abdominal thickness is smaller than 70 mm, a lifting mechanism in the

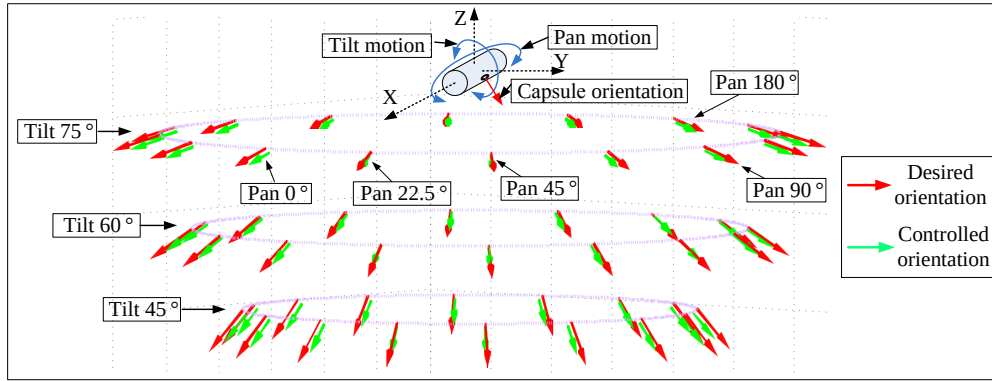


Figure 2.21: Tilt and pan motion experiment results under an abdominal wall thickness as 40 mm.

control unit can be developed to increase the capsule-to-control unit distance to the normal working range.

It is noteworthy to further consider the magnetic field effect on other surgical tools during laparoscopy surgery. Even though the generated magnetic field provides enough force for fixation and manipulation of the camera capsule in the range of 20 mm to 50 mm, the magnetic field drastically decreases in further distances from external control unit, due to its nature. As shown in Fig. 2.1, independent manipulation of camera capsule with respect to other instruments, reduces the possibility of interference with other instruments.

In order to insert/remove the camera capsule into/from the abdominal cavity through a single incision, a small continuum manipulator, which is capable of manipulating objects through an incision, will be designed with a gripper and a camera installed at the tip. Due to the strong magnetic fields surrounded by the capsule and the control unit, the material of the continuum manipulator should be selected as non-ferrous metals or plastic materials to avoid the magnetic field influence. To guarantee the continuum manipulator can provide sufficient force for placing and retrieving the robot, a minimum 30 g force should be considered to design the manipulator. In case of sudden removal of control unit and capsule fall inside abdominal cavity consequently, the continuum manipulator's gripper can be used for picking up the capsule. The manipulating feature of this system, eases the removal by controlling the orientation of camera capsule through the trocar/incision.

2.7 Summary

In this chapter, a closed-loop control system of a novel insertable laparoscopic camera has been presented to enable 2D orientation control, which consists of tilt motion control and pan motion control. The magnetic actuation mechanism features a unified control of anchoring, navigating, and rotating the camera without on-board electric motors. The tilt motion control was achieved by developing a control unit magnetic field function, an abdominal wall sensing system, and a cEPM adjusting mechanism with 0.5 mm resolution to generate optimal magnetic field from the EPMs. A closed-loop control scheme was presented to control current inputs of the coils in the control unit for desired tilt angles of the camera.

The pan motion control was achieved by designing a pan motion mechanism in the control unit to magnetically actuate the capsule. The experimental investigations indicate that our camera actuation mechanism design can achieve 0.594° and 0.524° average control errors in tilt motion control and pan motion control respectively.

In our future work, the camera on-board electronics will be integrated especially an inertial sensor, which is used to provide the camera orientation feedback wirelessly. The dimensions and the weight of the control unit will be further reduced for convenient usage by surgeons. The control unit cables will be removed by integrating a wireless module, coil drivers, and batteries inside the control unit. The camera system will be further tested *in-vivo* within a porcine abdominal cavity to verify the proposed control method and estimation methods. We will also further investigate the interference of generated magnetic field with other surgical instruments to minimize the disturbance. A proper solution for reducing the size and weight of external control unit would be replacing electromagnetic coils with permanent magnets. However, controlling the magnetic field generated solely with permanent magnets should be further investigated.

Chapter 3

Modeling and Analysis of a Laparoscopic Camera's Interaction with Abdomen Tissue

3.1 Abstract

Robotic camera systems have recently drawn attention in minimally invasive surgeries. Control and manipulation of these systems during traversing abdominal cavity is associated with camera-tissue interaction. This chapter demonstrates a theoretical and experimental analysis of a wireless laparoscopic camera's interaction with abdominal wall during a minimally invasive surgery. A mechanical model is developed in order to represent the behavior of abdominal wall bulk tissue by considering skin, fat, muscle and connective tissues layers. Model would predict the non-linear, viscoelastic behavior of the abdominal bulk tissue during interaction with a cylindrical laparoscopic camera. Model was implemented in ABAQUS to analyze the camera-tissue interaction and find the interaction forces generated during camera's contact and motion with different linear and rotational velocities. Simulations were validated by experiments on porcine tissue which can be used for proper control of insertable camera during a laparoscopy surgery. A noninvasive online method is proposed to measure the mechanical properties of each patient's abdominal wall tissue before start of MIS in order to tune control system to optimize the interaction depth and opposite forces. This method features preventing damage due to overload and camera fall during surgery.

3.2 Introduction

Single incision laparoscopic surgery (SILS) and natural orifice transluminal endoscopic surgery (NOTES) recently have drawn attention, since they can reduce the surgical incisions to minimum and recovery difficulties at post-operative stage. [84]. An important challenge in SILS and NOTES is the shared use of incision for both surgical instruments and camera trocar. This will make the control of surgical tools more complex and as a result the dexterity will be degraded. Also the quality of video feedback is decreased, since it may not be able to cover the proper view angles needed for the surgery by trocar pushing.

Different robotic systems have been developed to improve the video quality in surgeries. Challenges of these designs are mostly the manipulation, control and fixation of the camera

robotic systems inside the abdominal cavity. Researches at university of Columbia developed laparoscopic camera which uses DC servo motors for pan and tilt motion separately[87, 35]. Both of these designs use suturing in order to fix them to abdominal wall. The researchers at the University of South Florida designed a wireless laparoscopic camera [10] that uses a needle to pierce through a patients abdominal wall for camera fixation, and is manipulated by two motors for pan and tilt motions. Using suture and piercing mainly will disable the camera system to be translated to the new position inside the abdomen.

Wheeled surgical camera systems have been proposed to enable the translation of camera robots to desired locations. An in-vivo wheeled camera robot have been designed to traverse the organs to reach the proper position for video providing [59, 60]. The drawback of wheeled camera robot systems is the poor, near-horizontal view angle during the surgery which decrease the quality of hand-eye coordination for the surgeon.

One possible improvement for these designs is the application of a magnetic driven wireless laparoscopic surgical systems which traverse abdominal wall ceiling and can provide proper view angle for video capturing [36, 65, 79, 45].

An important drawback of most reviewed studies is ignoring the interaction of the surgical robot and soft tissues. Due to the complex characteristics of human tissues and its unique behavior under manipulations such as indentation, rotation and translation. An interaction study is crucial to predict the motion of camera robots inside the abdominal cavity for motion control and manipulation [25]. Another disadvantage of current systems is that most of them concentrate on homogeneous tissues like liver. while magnetic driven systems, traverse abdomen tissue which shows different properties based on patients physical condition and surgical area.

3.2.1 Problem Description of Insertable Cameras

A novel insertable magnetic driven robotic capsule camera system for single incision laparoscopic surgery was designed by Liu et al [43]. External stator controls anchoring, translation and rotation of an insertable robotic capsule camera (Rotor) by adjustment of magnetic field. As conceptually shown in Fig. 3.1(a),(b), the cylindrical camera is inserted into a gas filled abdominal cavity. The designed wireless laparoscopy camera system features

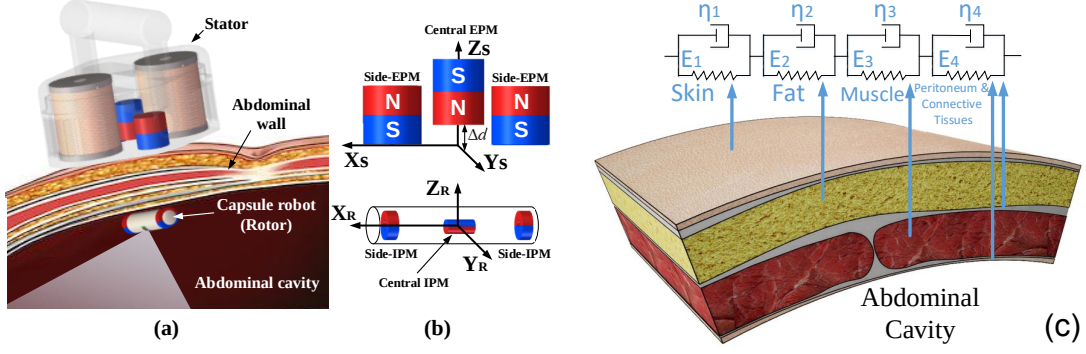


Figure 3.1: Wireless insertable capsule camera system. (a) Conceptual illustration of the wireless capsule robot. A stator is placed outside of an insufflated abdominal wall. The capsule robot is anchored, navigated and oriented by the stator to visualize the target surgical area. (b) Configuration of Magnets in stator and camera rotor. (c) Mechanical modeling of Abdominal Tissue with four Voigt models in series.

traversing the ceiling of abdominal cavity during surgery in order to illuminate the target surgical area and provide proper view angle and video feedback to surgeons.

Patient's abdominal wall thickness differs and has its specific properties based on the fat thickness, Muscle density, skin features and other physiological parameters. Since the system works based on magnetic field, the distance of stator to rotor is an important factor in controlling the camera capsule. This distance depends on patient's abdominal wall thickness which effects the mechanical properties of the tissue and changes its behavior during interaction with camera robot. Over-increasing this distance will diminish the anchoring force over the camera capsule and may cause the camera to fall inside the abdominal cavity which is undesirable. Another critical issue in magnetically driven surgery robots is the damage prevention which may arise from intense compressive force between stator and rotor during traversing tissue to the desired position.

These issues persuades to design a system capable of measuring each patients tissue properties online and in a non-invasive way. This data will be used to estimate the contact forces/torques and finally tune the stator's control system. The first step to reach this goal would be study of biomechanical properties of abdominal wall tissue. In general, Soft tissues are highly nonlinear, inhomogeneous, anisotropic, viscoelastic material and the modeling is of high importance not only in camera control but also in magnetic field

force/torque calculations. In order to successfully model the interaction between the camera and abdominal organ tissue, a constitutive material model is needed.

In this chapter, a mechanical model is proposed to model the bulk abdominal wall tissue by considering four voigt models. the creep experiment were conducted to find the model parameters. Model is validated through creep test on porcine abdomen tissue to represent the real tissue and then implemented in ABAQUS for finite element analysis of the Camera-tissue interaction. A mathematical model in order to study the wireless insertable camera-abdominal wall interaction is studied. An experiment setup is designed to derive the mechanical properties of porcine abdominal wall tissue. Interaction Experiments were designed to validated the simulations. These interaction forces are the key element in proper control of camera during a laparoscopy surgery.

The contributions of this chapter are: 1) A Biomechanical model of Bulk abdominal wall tissue is developed to simulate the abdominal wall behavior accurately by considering all tissue layers. 2) FEM Simulation of wireless insertable capsule camera's interaction with Abdominal wall during motion is presented and the interaction forces and torques are validated with experiments on excised porcine abdominal wall tissue for different motion speeds, normal loads and abdomen thicknesses. 3) Development of a control system for damage prevention during MIS and also for fall prevention during camera repositioning in abdominal cavity. 4) propose a method for noninvasive, online measurement of biomechanical properties of abdomen tissue during surgery based on the developed model. This method can be used for camera control by estimating the interaction forces online and tuning the stator's magnetic control system.

3.3 Mechanical Modeling of Abdominal Wall Tissue

A mechanical model is developed to study the bulk abdomen tissue. This model features considering all the major layers of the tissue. Considering the model to do online properties measurement, This model should be simple enough to decrease the simulation complexity, however accurate enough to represent the tissue behavior properly. The accuracy of this model is evaluated by experiment on porcine abdomen tissue specimen.

3.3.1 Abdomen Bulk Tissue Model

Living tissues' mechanical behavior has been studied widely, they are considered as viscoelastic, highly nonlinear, inhomogeneous and anisotropic materials. These properties will cause a complicated behavior of tissues during interaction. Accurate interaction study of a rigid camera capsule robot and abdominal wall tissue requires a comprehensive constitutive material model to represent the behavior of organ during the contact.

A mechanical, viscoelastic model of soft tissues has been proposed in [27] to simulate the creep behavior of viscoelastic materials. In this study, since the Camera-Tissue interaction during motion in surgery happens in short time ranges, four voigt models in series are used. The proposed model, shown in Fig. 3.1(c), can simulate the creep behavior of abdominal wall tissue. These four voigt models, each demonstrates a main part of the tissue which are skin, fat, muscle and the connective tissues. The effects of blood vessels and minor layers are also considered in connective tissue model.

In this mechanical model, when a constant load σ_0 is applied, the initial displacement will be caused by the springs with moduli E_1, E_2, E_3, E_4 shown in Eq. (3.1). Compression of dashpots with viscosities of η_1, η_2, η_3 and η_4 will result in following gradual displacements. Summation of these two displacements results in total displacement of tissue under uniaxial load. The total displacement ε of the model can be shown as

$$\begin{aligned} \varepsilon(t) = & \frac{\sigma_0}{E_1} (1 - e^{(\frac{-E_1 t}{\eta_1})}) + \frac{\sigma_0}{E_2} (1 - e^{(\frac{-E_2 t}{\eta_2})}) + \\ & \frac{\sigma_0}{E_3} (1 - e^{(\frac{-E_3 t}{\eta_3})}) + \frac{\sigma_0}{E_4} (1 - e^{(\frac{-E_4 t}{\eta_4})}) \end{aligned} \quad (3.1)$$

Parameters of this model can be fitted by creep experiment and measurement of the abdominal wall tissue specimen's displacement versus time.

3.3.2 Creep Test

5 specimens of porcine abdominal wall with cross section of 20 mm by 20 mm were cut and were used for creep test. Normal forces of 0.5, 1, 2, 3, 4, 5, 6 N were exerted over the specimens by the balance setup introduced in section 5.6 and their axial displacement were measured by the dial indicator. A camera image capture was used to record the dial

indicator measurements by rate of 25 fps. The lateral Displacement was also recorded by a camera and lateral strain was calculated by an image processing algorithm with resolution of 0.1 *mm*. Each specimen were tested four times. Results are shown in Fig.3.2 (b) and used to analyze the mechanical properties of the tissue discussed in section 3.3.

Data Collection and Parameter Fitting

The creep experiment setup, described in section 5.6, was used to collect the creep test data from abdominal tissue specimen with different loads. Vertical displacements were measured with dial indicator and the measurements were recorded by a 150 fps Camera(XIMEA xiQ). The normal load represents the compressive contact force between the camera robot and tissue during surgery. This contact force varies by the camera-stator distance which depends on the patient's abdominal wall thickness. Creep test load's range was selected between 0.5*N* and 6*N* to cover the probable contact forces' range during surgery. Each load's test was repeated 5 times and the average creep results are shown in Fig. 3.2 (a).

An iterative algorithm was used on this raw data to fit the parameters of Equation(3.1). The proposed four element voigt model parameters, E_1 , E_2 , E_3 , E_4 , η_1 , η_2 , η_3 and η_4 are summarized in Table 3.1 for each experiment.

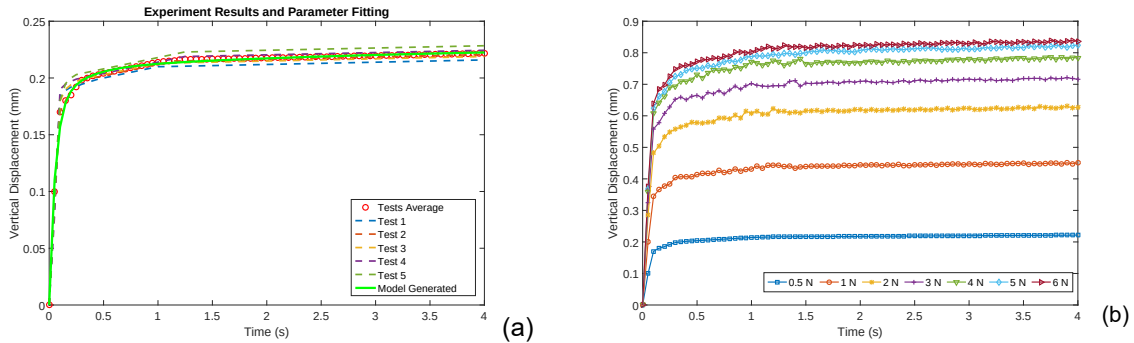


Figure 3.2: (a) Uniaxial creep test data of 0.5 *N* vertical force on porcine abdominal wall specimen. The dashed lines shows 5 test sets. The red circles shows the average of tests and Solid green line is model generated with parameters fitting. (b) Creep Results.

Table 3.1: Creep Curve fitted Parameters used in the simulation of creep behavior in Porcine specimen

Load (N)	Stiffness (kNm^{-1})				Dashpot viscosity (s)			
σ_0	E_1	E_2	E_3	E_4	η_1	η_2	η_3	η_4
0.5	15.5541	14.6631	14.0707	40.3930	0.6054	1.3378	1.1773	107.0078
1	13.1737	5.8432	5.7823	9.9118	0.4058	0.4828	0.4779	99.9246
2	13.0266	3.8522	3.8664	9.8459	0.3479	0.3296	0.3308	99.8941
3	12.7450	3.3139	3.2974	9.8135	0.3322	0.2834	0.2821	99.8837
4	12.7865	2.9956	2.9798	9.8182	0.2969	0.2560	0.2547	99.8876
5	8.0462	2.5984	3.6073	9.8619	0.3518	0.1083	0.7728	99.7520
6	7.4735	2.7467	3.2855	9.8649	0.3349	0.1135	0.5987	99.9762

3.4 Camera-Tissue Interaction Mathematical Model

A mathematical model of Capsule camera-Tissue contact is studied and the equations of motions for vertical and horizontal motions were derived. This model can be used for online estimation of interaction depth in control system. A mathematical model for interaction of wheeled mobile robot and liver tissue was proposed in [58]. The same procedure has been applied here, but the abdominal wall tissue is considered as a generalized kelvin-voigt model for rotation-less translation of camera capsule, since the actuation is provided through magnetic coupling.

The free body diagram of the camera capsule during interaction with tissue is shown in Fig. 3.3 (a), where F_w shows the gravity force over the capsule. The contact force is shown by F_c which is resolved on x and y axes as F_{cx} and F_{cy} respectively. The magnetic force exerted over the camera is represented as F_m and its components are shown as F_{mx} and F_{my} .

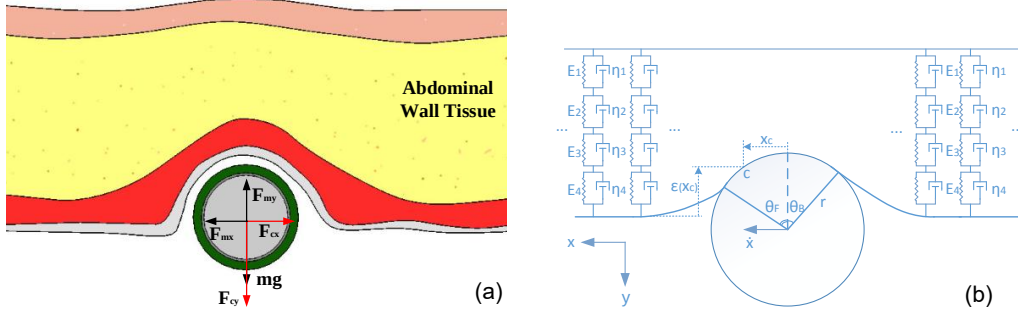


Figure 3.3: (a) Free body diagram of the camera capsule in contact with abdomen (b) Schematic view of camera-Tissue contact with 4 voigt model.

The contact force is generated from different sources during interaction. A simple model has been considered which consists of viscous force of peritoneum fluid which covers the organ and stress from viscoelastic nature of the organ. Due to the small effect of tissue membrane tension it was not considered in this model.

$$\mathbf{F}_c = \mathbf{F}_{viscous} + \mathbf{F}_{stress} \quad (3.2)$$

The stress originated from the Displacement of the tissue and because of asymmetry of front and back contact lengths, the resultant stress would be non-zero.

Displacement profile of the membrane in front of the rolling wheel is approximated as a decaying exponential function in the x -direction and also for rolling motion, in the viscoelastic model, the length of contact behind of the rolling cylinder is less than front of it due to the viscoelastic nature of the tissue.[58]. These assumptions are used for the linear motion modeling and the schematic view of contact is shown in Fig. 3.3 (b) and the Displacement profile is assumed as:

$$\varepsilon(x_c) = Ae^{B(x_c-C)} \quad (3.3)$$

where the constant C is the length of contact along x -direction, $C = r\sin(\theta_F)$ and r is the radius of capsule camera. By considering the front contact point at $x_c = r\sin(\theta_F)$ and the capsule geometry, the constant A , B are found. The same procedure can be solved for the backside contact point and the Displacement profiles can be proposed as

$$\varepsilon(x_c) = \begin{cases} r\cos(\theta_F - Y_{center})e^{-\frac{\tan(\theta_F)}{r\cos(\theta_F - Y_{center})}(X_c - r\sin(\theta_F))} & \text{Front} \\ r\cos(\theta_B - Y_{center})e^{-\frac{\tan(\theta_B)}{r\cos(\theta_B - Y_{center})}(X_c - r\sin(\theta_B))} & \text{Back} \end{cases} \quad (3.4)$$

where Y_{center} is the position of capsule's center of mass on Y -direction.

Finding equations of motions requires to consider the forces acting on the camera capsule along each axis, which are the shear force and the tissue pressure. The shear force generated from the viscous peritoneal fluid covered the internal organs. Due to asymmetry of contact lengths in front and backside, the shear stress generates force on both x and y directions. by projecting the shear force along these axis the generated force can be estimated.

$$F_{Viscous} = \begin{cases} L \int_{\theta_B}^{\theta_A} \tau \cos(\theta) r d\theta & x - direction \\ L \int_{\theta_B}^{\theta_A} \tau \sin(\theta) r d\theta & y - direction \end{cases} \quad (3.5)$$

Where L is the length of camera capsule and τ is shear stress at the capsule interaction which can be modeled as a Newtonian fluid with linear velocity profile in contact region

$$\tau = \mu \frac{-\dot{x}\cos(\theta)}{h} \quad (3.6)$$

where μ is the peritoneal fluid viscosity and $\dot{x}\cos(\theta)$ is the relative linear velocity between camera capsule and the tissue and h is normal distance between these two. μ and h can be calculated experimentally.

The pressure force can be estimated from the Displacement of the tissue in contact with capsule camera as follows

$$F_{Stress} = \begin{cases} -L \int_{\theta_B}^{\theta_A} r \sigma(\theta) \tan \theta d\theta & x - direction \\ L \int_{C_B}^{C_A} \sigma(x_c) dx_c & y - direction \end{cases} \quad (3.7)$$

where $C_B = r\theta_B$ and $C_F = r\theta_F$ are the back and front contact lengths respectively. Also $\sigma(x_c)$ can be found from the viscoelastic response of the tissue:

$$\sigma(x_c) = \frac{\varepsilon(x_c)}{\sum_{i=1}^4 \frac{1}{E_i} (1 - e^{-\frac{E_i}{\eta_i} t})} \quad (3.8)$$

The equations of motion can be formulated as

$$\begin{cases} m\ddot{x} = F_{mx} + F_{viscous-x} + F_{Tissue_x} & x - direction \\ m\ddot{y} = mg + F_{viscous-y} + F_{Tissue_y} - F_{my} & y - direction \end{cases} \quad (3.9)$$

where m is the mass of the capsule camera.

3.5 Interaction's Finite Element Simulation

Interaction of camera capsule and abdomen tissue was simulated in ABAQUS 6.14 to study the contact forces and deformations of the tissue during linear motion and pan motion. First, the creep test was simulated to validate the implemented model. Next, the static and dynamic model of interaction was simulated for different interaction forces and motion speeds.

3.5.1 Viscoelastic Model Validation

The derived model in section 3.3 was simulated in ABAQUS to compare the results of the experimental creep test with the simulation. The abdominal wall tissue material was simulated as a four layer viscoelastic model in ABAQUS by the properties derived in section 3.3. The properties of each layer was implemented into ABAQUS to replicate porcine tissue. The comparison of creep experiment and Creep simulation shown in Fig.3.3 (b) reveals less than 5% deviation. It's concluded that the derived model can estimate the abdominal wall tissue response properly and can be used for studying the Interaction.

3.5.2 Camera-Tissue Interaction

The capsule camera with the length of 80 mm and weight of 50 gr is modeled in ABAQUS as an analytical rigid body with 1600 elements. For simulating the abdominal wall tissue, a cylindrical viscoelastic model with radius of 150mm, thickness of 30mm and 30000 elements was considered. Fixed boundary conditions are considered for the model to replicate the experiment. Different Normal forces were exerted over capsule to represent the magnetic

force in real experiment which compress the capsule toward the abdominal wall. Friction coefficients were considered as 0.1 for static and 0.075 for dynamic and the gravity was also considered.

Two sets of simulation were done to represent the linear motion and pan motion separately. the simulation divided into two steps, first the gravity and normal concentrated forces are exerted over the capsule which can be seen in Fig.3.4 (b). Time $t = 0$ shows the indentation only without any motion. Second step will simulate the motion of the capsule camera during the contact. Fig.3.4 (c) shows the stresses generated over the tissue during linear motion with constant speed of 10 mm/s . Fig.3.4 (d) shows the contact stresses after 1.776 s during a pan motion with constant angular speed of $\pi/4 \text{ rad/s}$.

The contact forces during these motions are exported from ABAQUS and will be used in comparison with experiment results in section 5.6. The horizontal force exerted over the camera capsule model during a linear speed motion with different linear speeds is shown in Fig.3.5 (a). After the first second, when the contact is established, the rotor starts to move in horizontal plane with different speeds. Increasing the linear speed results in reduction of contact force. This effect can be justified by the fact that increasing the speed, will reduce the contact depth and will result in less contact force.

The same simulations were run for different speeds of pan motions in order to find the generated torque during constant speed pan motion of camera robot. Angular velocities were

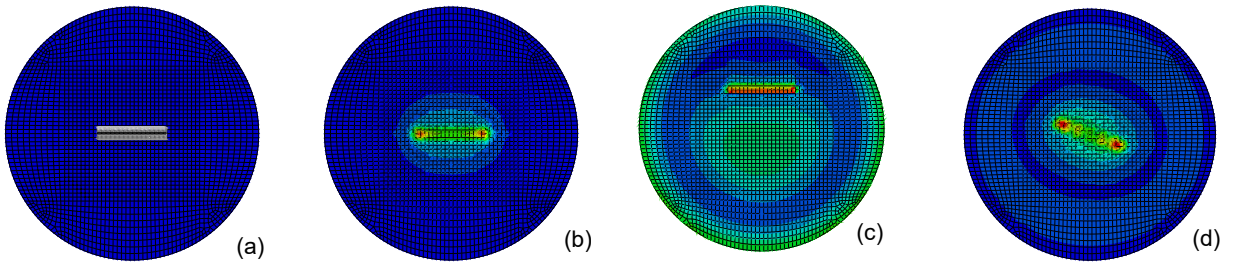


Figure 3.4: Finite Element Simulations of Interactions: (a) Initial state of camera capsule and Tissue without any contact. (b) Initial contact of camera capsule and Tissue when only normal force is exerted over the tissue. (c) Contact of camera capsule and Tissue during a linear motion with speed of $v = 5 \text{ cm/s}$ at time $t = 1.6 \text{ s}$. (d) Contact of camera capsule and Tissue during a pan motion at time $t = 1 \text{ s}$ with the angular velocity of $\omega = \pi/4 \text{ rad/s}$.

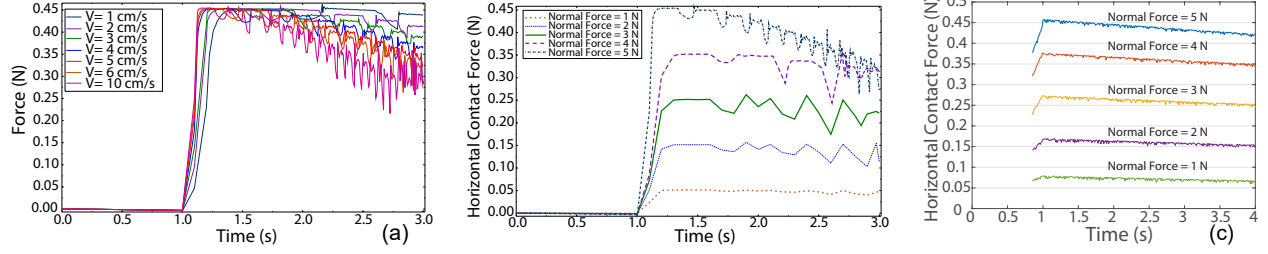


Figure 3.5: FEM Simulations and experiments of linear motion interactions: (a) Horizontal contact force generated during linear motion of camera with different linear speeds in ABAQUS with Normal force of $5N$. The first second, is the indentation step and the horizontal contact force is zero. (b) Different normal forces were exerted over capsule camera and generated horizontal contact force were simulated during a linear motion of $v = 5 \text{ cm/s}$. (c) Experiment result for measuring generated horizontal force with a linear motion of $v = 5 \text{ cm/s}$ under different normal forces. The first 1 second is considered for proper indentation of camera capsule over the tissue and measurements are started at $t = 0.85 \text{ s}$.

chosen in the range of $\omega = \pi/6$ to $\omega = \pi \text{ rad/s}$ to cover the probable actuated pan motion of ex-vivo stator.

3.5.3 Different Normal Forces vs Contact Force

Based on the thickness of patient's abdominal wall, distance of stator to rotor would change and consequently, the normal attractive force between magnets would vary. Different normal interaction forces of the camera capsule and the contact were simulated for a case of linear speed $v = 5 \text{ cm/s}$. Generated horizontal force shows that by increasing the normal contact force, the generated lateral force would increase which is a rational result. Since increasing the normal force would increase the friction force between rigid capsule and soft tissue, also the depth of the indentation would increase which leads to more resistant force.

3.6 Experimental Evaluation

In this section the design of experiments conducted will be discussed. All the experiments are done on the basis of the balance setup. This balance setup has sub-gram resolution in weight measuring, 0.0254 mm resolution in displacement measurements and has the feature

to cancel the indicator spring force for the experiments. The two dial indicators also keep a constant-load contact during the motion of tissue below the camera capsule.

3.6.1 Tissue Specimen

For the experiment excised porcine abdominal wall was used. 20 mm by 20 mm cross section specimens were cut for creep test. Each specimen contains different layers of skin, Fat, Muscle and connective tissues. A 150 mm by 150 mm cross section of abdominal wall was used for measuring the contact forces during linear motion and pan motion of capsule camera over the tissue.

3.6.2 Experimental Setup

A balance setup was designed in order to run the experiment with high accuracy in measuring the normal Displacement and the normal force exerted over the tissue. For this purpose two identical Dial indicators(All Industrial Inc) with the resolution of 0.0254 mm were used on the two opposite sides of the balance setup. This design as shown in Fig.3.6 (a), will help to cancel the stiffness of dial indicator's stiffness. since the range of Displacement is less than 3 mm , by pre-adjusting both indicators to 10 mm , it can be assumed that the springs are linear and can cancel each others force during measurements. This design also helps to keep the contact during the motion and overcome the oscillations. This setup is sensitive to sub-gram loadings and is used as base for creep and interaction force experiments.

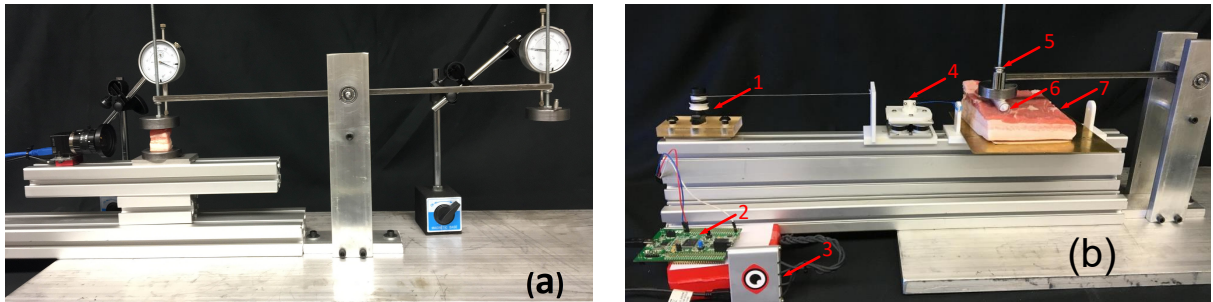


Figure 3.6: (a) Balance setup. (b) Linear motion experiment to measure lateral interaction force. Components of setup: 1-Servo Motor 2-STM32 MicroController 3-OptoForce Force Sensor 4-Force sensor DAQ 5-Normal Load weights 6-Camera capsule 7-Porcine abdominal wall tissue.

3.6.3 Interaction Force Test

A set of experiments were done to find the effect of normal force on generated contact force during a constant speed linear motion. The Normal force range was selected as 1 N to 5 N with a constant speed of $v = 5 \text{ cm/s}$. The results of this experiment are shown in Fig.3.5 (c) and comparison with the ABAQUS simulation analysis results shows a good compliance.

3.6.4 Linear Motion Interaction Force Detection

Purpose of this experiment is to find the required horizontal force which should be exerted on camera capsule to move with constant linear speed during interaction with soft tissue. This force can be used to estimate the force generated by interaction during traversing of abdominal wall. For this experiment the balance setup was used and a constant mass of 200 gr were used to exert normal force over the camera capsule over the tissue while Camera capsule was fixed to the balance setup. The tissue below the camera was driven by a servo motor at constant speed of 1 cm/s and the horizontal force created by interaction was measured by a six axis force/torque(F/T)sensor (HEX-58-RB-2000N, OptoForce Inc.) as shown in Fig.3.6 (b).

The force sensor and the tissue were aligned over a rail to make sure only the force generated by contact along this line was measured. The friction force between the carts holding sensor and tissue was canceled by repeating contact-less test.

3.6.5 Pan Motion Interaction Force Detection

The same procedure as linear motion experiment was used for pan motion Experiment as shown in Fig.3.7 (a). The Tissue was put over a rotating disk with the diameter of 240 mm while the camera capsule was pushed toward the tissue by a constant normal load, generated by balance setup. A servo motor was pulling the force sensor over the bar with constant speed and generating constant angular velocity rotation of disk. Different angular velocities with the range of $\omega = \pi/6$ to $\omega = \pi \text{ rad/s}$ were tested to measure the torque generated during contact. The sensing force multiplied by rotation radius, reveals the Generated torque

during contact. A comparison of Simulation and Experiment results are shown in Fig.3.7 (b) which reveals good compliance.

3.7 Damage Prevention, Fall prevention and Online Tissue properties Measurement system

A stator-Rotor Distance estimation mechanism was designed in our previous work [45]. The main idea of estimating stator-rotor distance h is to sense the rotor magnetic field by using a pre-built rotor magnetic field map which is with respect to different h . To sense the magnetic field from the rotor, four sets of tri-axis hall effective sensors were installed at the bottom of the stator. Different parameters of this mechanism are defined in Fig.3.8 (b). D is the distance between stator and patient's stomach and can be adjusted by an arm robot, controlling the stator's position and orientation.

Due to the lack of haptic feedback in robot-assisted MIS, tool-tissue overload may result in collateral tissue damage. in order to minimize tissue damage in MIS, a force control strategy seems crucial. By knowing damage thresholds for abdomen tissue, it can be incorporated into the control scheme so that instrument forces are limited to this level and tissue overload is actively prevented [19].

Two miniature force sensors (Tekscan FlexiForce A101) are attached to the external shell of camera capsule to measure the contact force (F_c) online during the tissue interaction. Fig.3.8 (c) illustrates the Camera-Tissue interaction force control architecture. The system

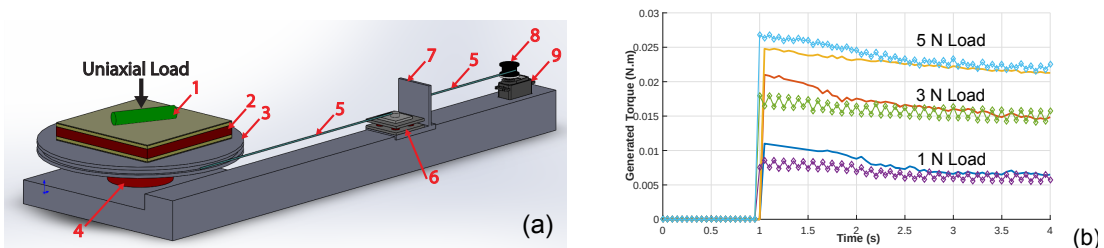


Figure 3.7: (a) Pan Experiment setup consists of : 1- Camera Robot. 2- Abdominal Wall tissue. 3- Rotating Disk. 4- Bearing. 5- Strings. 6- Force Sensor. 7- Sliding Base of Force sensor. 8- Pulley. 9- Servo Motor. (b) Pan experiment results(Solid Line) and comparison with ABAQUS Simulations (Diamond Line) for different forces of 1N, 3N, 5N with $\omega = \pi/2$

initially measures the distance of stator to rotor (h) and the sensed contact force will be compared to the desired contact force. Deviation from desired force will cause the stator-holder arm robot to adjust the stator-rotor distance (D) Fig.3.8 (a).

The benefits of contact force control system are: 1) During the surgery it will reduce the possibility of collateral injuries caused by force overload. Since the abdominal wall thickness is specific for each patient and also each section of abdomen has particular thickness, this system can control the contact force online during the surgery while the camera is traversing abdominal cavity ceiling. 2) A very important aspect in a magnetic driven camera system would be fall preventing assurance. Excessive increase in stator-rotor distance due to the abdominal wall thickness, may lead to loss of anchoring force on camera and consequently camera falling inside the abdominal cavity. This control system can prevent the falling by limiting the stator-rotor distance based on sensed force. The control system will always keep (h) in the range where $F_{min} < F_{-c} < F_{max}$.

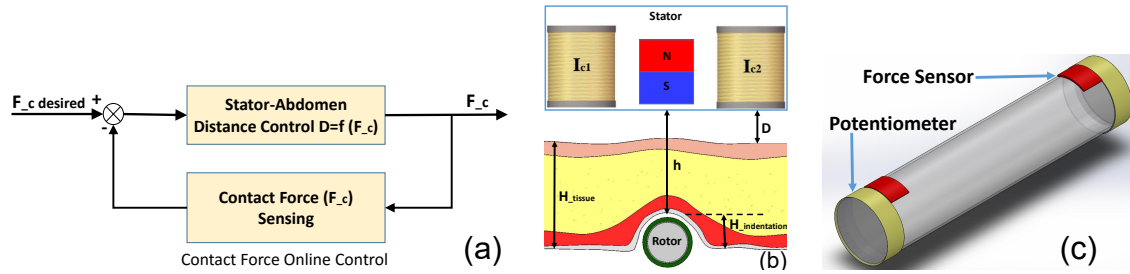


Figure 3.8: (a) Online Camera-Tissue interaction Force Control Strategy. (b) Force estimation mechanism. h is the stator-Rotor distance estimated by hall effective sensors. D is distance of stator and patient's stomach. H_{tissue} is abdomen's thickness. $H_{indentation}$ is the camera's indentation depth caused by magnetic pull. (c) Camera capsule with force sensor to sense contact normal force and potentiometer to sense indentation depth.

3.7.1 Online Tissue Properties Measurement

Patient's physical condition and obesity can influence the magnetic driven system's performance and control. Abdominal wall thickness features has influence on the properties of tissue. These properties are different for each patient and even different for different abdomen areas, As a result the camera-tissue interaction depth and forces will be variable for different patients and abdomen areas. In order to control the camera inside the abdominal

cavity, interaction forces should be estimated online and being feedback to the control system. Finding the interaction forces requires an online tissue properties measurement system.

A method has been developed here in order to measure the properties of tissue online. The main idea is to regenerate the creep test in-vivo with the camera and stator. The system measures the indentation depth $H_{\text{indentation}}$ during the interaction by Two ribbon potentiometers attached over the camera capsule. The resistance measured by potentiometers demonstrates the length of contact. As shown in Fig.3.3 the length of contact can be shown as $r(\theta_F + \theta_B)$. Since the online creep test is done in static state, $\theta_F = \theta_B$. By using the exponential profile defined in Eq.3.3, The real indentation depth can be measured. Online properties measurement is done at the beginning of camera insertion and the indentation data versus time gives the creep response of specific patient's tissue. The creep test data will be implemented in the mechanical model of section 3.3 to generate the contact forces and the control system will be tuned.

3.8 Summary

In this chapter Interaction of a wireless insertable laparoscopy camera and abdominal wall tissue during motion is studied. For an accurate study, first an accurate mechanical model of abdominal wall tissue was required to represent the complex behavior of alive tissues during interaction. Therefore, a bulk model considering all tissue layers is proposed. The creep experiment helped to find the parameters of the model and finite element analysis confirmed the model. The camera system we studied, features pan and linear motion, so these motions were simulated in ABAQUS to find the interaction forces. Comparisons show good compliance between simulation and experiment results. A smooth motion and accurate control of camera capsule is possible only when these interaction forces are known during traversing in surgery. These results were validated by linear and pan motion interaction experiments designed over a bulk porcine abdominal wall.

An online noninvasive property measurement system is suggested which can use this model. This system will provide the required data for stator's control system for each specific patient during surgery. This system also prevents the force overload over the tissue and can

cancel camera fall during the MIS which are critical issues in magnetic driven systems. In our future work, a closed loop control system will be developed to control the camera motion during interaction based on the normal and horizontal contact forces. The Insertable wireless camera system will be used for laparoscopy surgery on alive pigs.

Chapter 4

Optimized Illumination for Superior Imaging Quality

4.1 Abstract

This chapter presents an initial prototype of an *in-vivo* robotic laparoscopic camera that features optimized illumination to address the problems, i.e. inferior lighting uniformity and low optical efficiency, in the state-of-the-art designs of *in-vivo* laparoscopic cameras. Benefiting from the transformable structure of the robotic camera, sufficient on-board space is created without sacrificing the camera’s compactness to carry three dedicatedly-designed freeform optical lenses for achieving the optimized illumination requirements. Designing miniature freeform optical lenses for extended light sources, such as LEDs, is a very challenging task which usually involves solving a non-standard Monge-Ampère equation. In this chapter, we approach the illumination optical design based on a ray-mapping method that is governed by a standard Monge-Ampère equation. We propose an effective numerical solver to compute the ray-mapping solution for constructing freeform lens surfaces. Experimental results prove the predicted performance of the illumination system design: greater than 97% illuminance uniformity, greater than 80% optical efficiency, and greater than 14.323 lx illuminance on a target plane with a distance of 100 mm. The effectiveness of this prototype is also experimentally verified by performing a suturing task in a simulated abdomen.

4.2 Introduction

In minimally invasive surgery (MIS), a laparoscope has been clinically adopted for visualizing the interior of the abdominal cavity since the mid-1980s [71]. A modern laparoscope system consists of one or more high definition (HD) digital cameras, an external light source such as xenon or light emitting diodes (LEDs) [11], a light transmission fiber-optic scope combined with other optical components for uniform illumination [88], a rod lens system for imaging, and a monitor. Superior laparoscopic imaging quality can thus be offered in terms of clarity and color definition. However, in the case of single-port access surgery, loss of triangulation between a laparoscope and other surgical instruments becomes the major disadvantage [57], which results in the clashing of instruments. In addition, counter-intuitive manipulation and limited field of view (FOV) of a laparoscope further degrade its performance in single-port access surgery. Alternatively, *in-vivo* laparoscopic cameras are developed that can be fully inserted into the abdominal cavity for providing dexterous manipulation without having the abdominal wall fulcrum constraint.

For designing *in-vivo* laparoscopic cameras, recent research studies have focused on developing robotic actuation mechanisms by magnetic couplings [40, 69, 42] and/or motor-driven articulated structures [34, 9] for pose control of the cameras, analyzing wireless transmission of imaging data and control signals via the abdominal wall [9], and clinically validating the effectiveness of *in-vivo* laparoscopic cameras [5, 74]. In spite of the achieved progresses, their inferior imaging quality is a major issue that obstructs the *in-vivo* camera designs to be clinically employed. It is not a straightforward task to obtain high quality images by only improving *in-vivo* imaging systems with HD imaging sensors. Because for the same sensor size, the higher resolution results in the smaller pixel size that is usually with the lower light sensitivity [20]. Sufficient light intensity within a camera's FOV is a necessity to enable HD imaging sensors to work in normal condition. In addition, non-uniform lighting distribution on a target visualized area results in poor quality images with shining centers and dim margins. Furthermore, light rays that emit from the on-board light sources should be projected within a camera's FOV to maximally utilize light energy. The

above requirements provide guidelines for designing an optimized illumination system in an *in-vivo* laparoscopic camera.

The illumination systems that are integrated in the state-of-the-art *in-vivo* laparoscopic cameras utilize bare LEDs [69, 34, 5] or LEDs stacked with reflectors [9] to realize basic illumination for the imaging sensors. To design an optimized illumination system, we need to address two major challenges: First, the robotic camera design should be compact enough to fit in the abdominal incision and meanwhile reserve sufficient on-board space to carry dedicatedly-designed secondary optics² of LEDs. Second, to satisfy the optimized illumination requirements, freeform non-imaging optics are needed to control light beams. Designing freeform optics is a challenging task because this problem is governed by a non-standard Monge-Ampère equation [66, 61], which is a second-order highly nonlinear partial differential equation (PDE). Moreover, the assumption of zero-étendue (point light source) will be invalid considering the comparable sizes of the LEDs and the freeform optics. This will make the design task even more difficult.

The scope of this work is to propose a solution to develop an *in-vivo* laparoscopic camera with an optimized illumination system. We intend to propel *in-vivo* laparoscopic cameras to take one step towards having comparable imaging quality with the up-to-date laparoscopes. In this chapter, we present an initial prototype of a transformable *in-vivo* robotic laparoscopic camera, as shown in Fig. 4.1, which features three foldable wings for creating sufficient space to contain three high-efficiency white LEDs as the light sources and three freeform optical lenses for controlling light beams. The folded mode in Fig. 4.1(a) is used to insert into the abdominal cavity via an incision, while the extended mode in Fig. 4.1(b) is activated to expose the imaging system and the illumination system after the robotic camera being anchored against the abdominal wall. The non-coaxial arrangement of the imaging system and the illumination system introduces shadow depth cues that help surgeons to perceive three-dimensional (3D) depth from two-dimensional (2D) images [31]. Freeform non-imaging optical lens design for the LEDs is one of the most critical tasks in this work. We approach this task based on a ray-mapping method which is governed by a standard Monge-Ampère

²Secondary optics are the optics which exist outside of an LED package, such as reflectors or lenses, to create the desired illumination appearance and beam pattern.

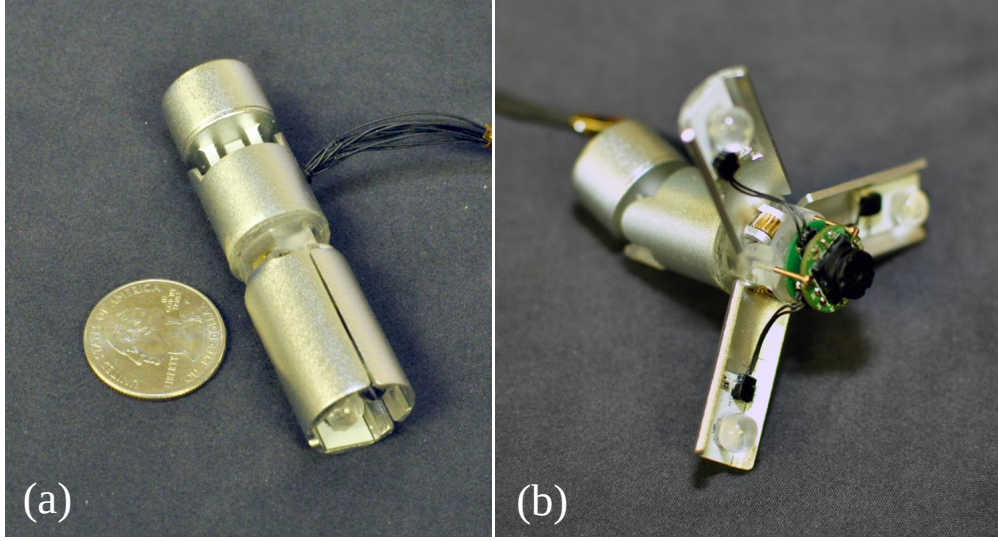


Figure 4.1: Prototype of the *in-vivo* robotic laparoscopic camera. (a) shows the folded mode of the camera, which is used for inserting into a trocar; (b) shows the working mode of the camera.

equation. We propose an effective numerical solver to compute the ray map solution for constructing freeform surfaces. To improve the degraded illumination performance that caused by extended light sources, a feedback modification scheme is introduced to adjust freeform surfaces.

The major contributions of this research include: (1) an innovative transformable robotic laparoscopic camera design, which combines compactness for fitting in the abdominal incision and sufficient on-board space for containing the optimized illumination system; (2) an effective freeform non-imaging lens design method for extended light sources, which is used for developing the optimized illumination system; (3) an experimental performance validation of an assembled illumination system with manufactured freeform lenses; and (4) a surgical task demonstration by a complete initial prototype that consists of an *in-vivo* robotic laparoscopic camera, an external anchoring/control unit, and a user controller for sending control signals.

4.3 Design of robotic laparoscopic camera system

4.3.1 System Overview

The *in-vivo* robotic laparoscopic camera system consists of an external anchoring/control unit (EACU), a robotic camera, a user controller, and a monitor, as shown in Fig. 4.2(a). During MIS, the robotic camera with the folded mode is inserted into the abdominal cavity via a trocar. The EACU magnetically fixes the robotic camera against the inner side of the abdominal wall. The soft cable between the EACU and the robotic camera is used for control signal transmission, imaging data acquisition, power supply, and retrieval of the robotic camera from the abdominal cavity.

A surgeon can send control signals to the microcontroller unit (MCU) in the EACU via the user controller, which controls open/close of the camera wings, pan/tilt motion to adjust the camera pose, on/off of the imaging system and the illumination system, and brightness of the LEDs.

4.3.2 External Anchoring/Control Unit (EACU)

The EACU is the central control unit of the whole system. Acquired surgical videos from the robotic camera is processed in the EACU and consequently send to the monitor in real time, as shown in Fig. 4.2(b). After receiving control commands from the user controller, the signals are processed in the MCU and transmitted to the actuator drivers, the LED driver, and the video processing unit for controlling the actuators, the LEDs, and the imaging sensor in the robotic camera.

The internal structure of the EACU is demonstrated in Fig. 4.3. Besides the controller circuit board, the EACU contains a diametrically magnetized external permanent magnet (EPM) that is rotationally driven by an actuator. The EPM is magnetically coupled with the robotic camera's internal permanent magnet (IPM) to provide anchoring force for the robotic camera's fixation against the abdominal wall, and spinning torque for pan motion control. The tilt motion of the robotic camera is driven by its on-board actuation mechanisms.

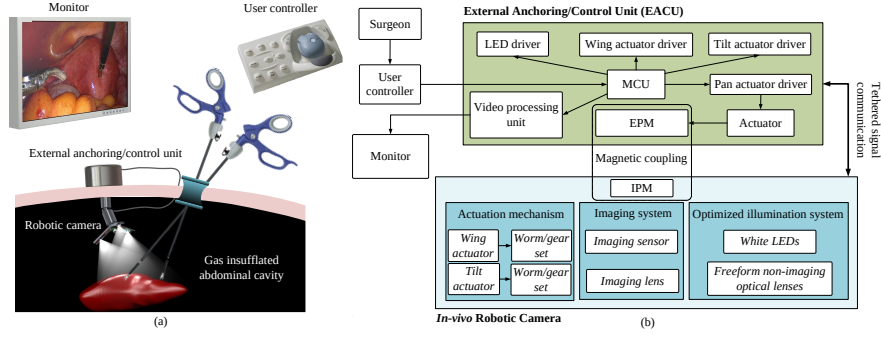


Figure 4.2: Functional diagram and physical configuration of the *in-vivo* robotic laparoscopic camera. (a) conceptually demonstrates the robotic camera system working in a surgical scenario; (b) shows the firmware design architecture of the robotic camera system.

4.3.3 In-Vivo Robotic Camera

The initial prototype of the *in-vivo* robotic camera currently consists of three main components: 1) an actuation mechanism that controls the camera's tilt motion and wings' opening angle; 2) an imaging system; and 3) an optimized illumination system (see Fig. 4.2(b)).

Actuation mechanism

In addition to the pan motion actuated by the EPM, the tilt motion of the robotic camera is controlled by an on-board actuator with a set of worm and gear, as shown in Fig. 4.4(a). The tilt gear is fixed with the shaft and the magnet head. The actuators' housing together with the camera and the illumination system rotates around the shaft by actuating the tilt worm. The combination of pan and tilt motion enables the robotic camera to visually cover the whole surgical area. The other actuator in the housing controls the wings' opening angle by using one worm and three gears. The actuation mechanism provides a tilt motion range of 49° and wings motion range of 80° (0° for the folded mode).

Fig. 4.4(b) shows the components of the actuation mechanism in the prototype. The actuators are stepper motors with diameters of 4 mm, lengths of 14.42 mm, and 125:1 planetary gearheads (ZWBMD004004-125, Shenzhen Zhaowei Machinery & Electronics Inc.). The stepper motor can deliver a torque of 10 mNm at continuous operation. The customized

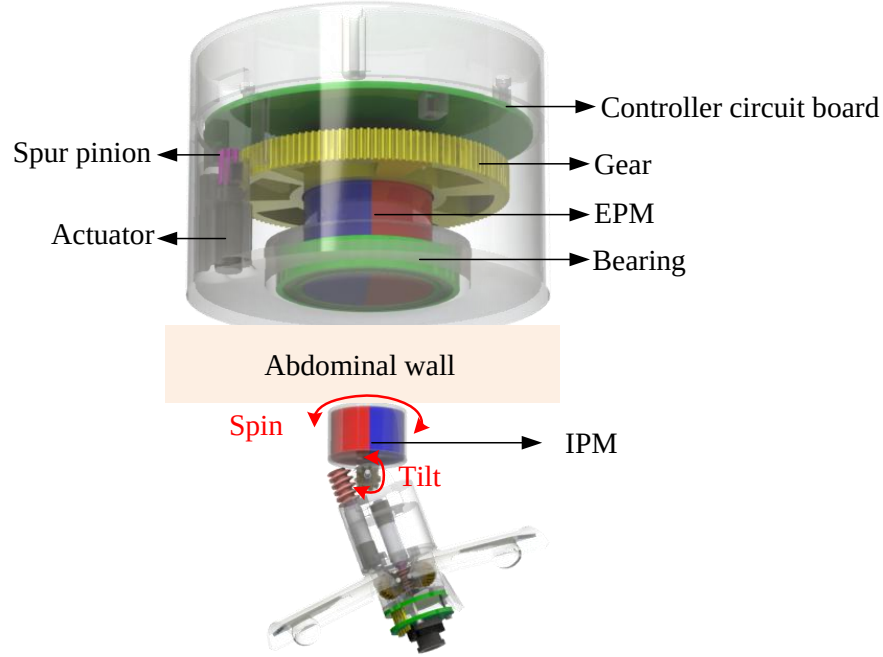


Figure 4.3: Demonstration of the EACU's internal structure and the robotic camera's pose control.

worm gear sets for the tilt actuator and the wings' actuator have reduction ratios of 12:1 and 20:1 respectively.

Imaging system

The imaging system in this initial prototype is developed to provide basic visualization functionality for evaluating the design performance of the *in-vivo* robotic camera system. Fig. 4.4(c) shows the developed imaging system that employs (1) an OmniVision OV7955 CMOS sensor with an imaging array size of 672×492 , an optical size of $1/3.7''$, a pixel size of $6 \mu\text{m} \times 6 \mu\text{m}$, a capture rate of 60 fps, and NTSC analog output; (2) a $1/4''$ lens (DSL756, Sunex, Inc.) with a focal length of 3.8 mm, a diagonal field of view 60° , and multi-megapixel resolution; and (3) a compatible lens holder (CMT756, Sunex, Inc.) for the DSL756 lens.

Optimized illumination system

As we discussed in Section 4.2, the objectives of designing an optimized illumination system are sufficient light intensity, high optical efficiency, and high illuminance uniformity. In fact, defining the "sufficient light intensity" for an imaging sensor is difficult due to many factors,

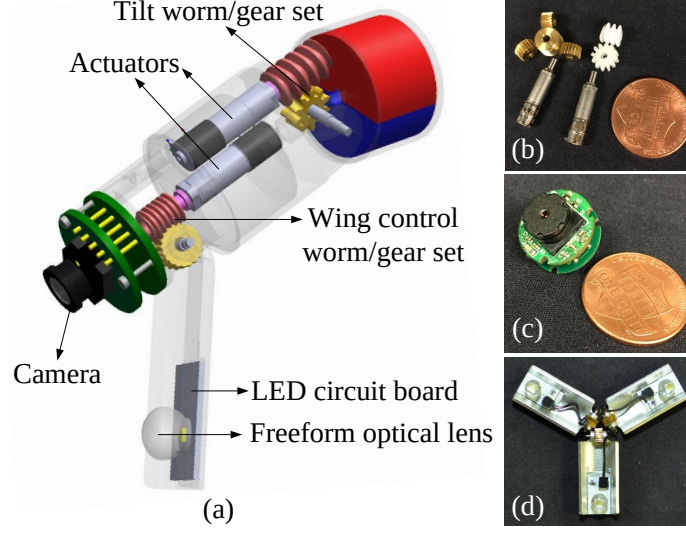


Figure 4.4: Internal structure of the robotic camera. (a) Structure overview. (b) Miniature actuators and worm/gear sets. (c) The camera module integrated with an imaging lens. (d) The illumination system with LEDs and freeform optical lenses.

e.g., pixels' sensitivity, surface conditions of an illuminated area, and the aperture size of an imaging lens. But according to some experimental studies [69], a maximum illuminance of 4.000 lx at a distance of 50 mm was used to work with a similar imaging sensor as the one in our imaging system. Considering 50 mm $\sim$ 100 mm working distance between the camera and a target area, we conservatively require our illumination system to uniformly deliver a minimum illuminance of 10.000 lx at a distance of 100 mm.

For maximizing light energy efficiency, Fig. 4.5(a) provides the feasible configuration that the camera's FOV fits right in the illuminated area. Due to the 60° diagonal FOV of the imaging lens, the radius of the illuminated area is calculated as $R = D \cdot \tan(60^\circ/2)$, where D is the camera-to-target distance. When $D = 100$ mm, R should be 57.7 mm. To guarantee the area of the camera's FOV can be uniformly illuminated, R is set as 80 mm considering design tolerance.

Fig. 4.4(d) shows the components in the optimized illumination system. Three white LEDs (Xlamp XQ-E, Cree, Inc.) with 5700 K color temperature are employed to provide a total luminous flux of 354 lm at 350 mA. If the luminous flux of 354 lm is uniformly projected on the area with $R = 80$ mm, an average illuminance of 17.607 lx can be achieved which is more than the required 10.000 lx. If we only use two LEDs, it seems that a total luminous

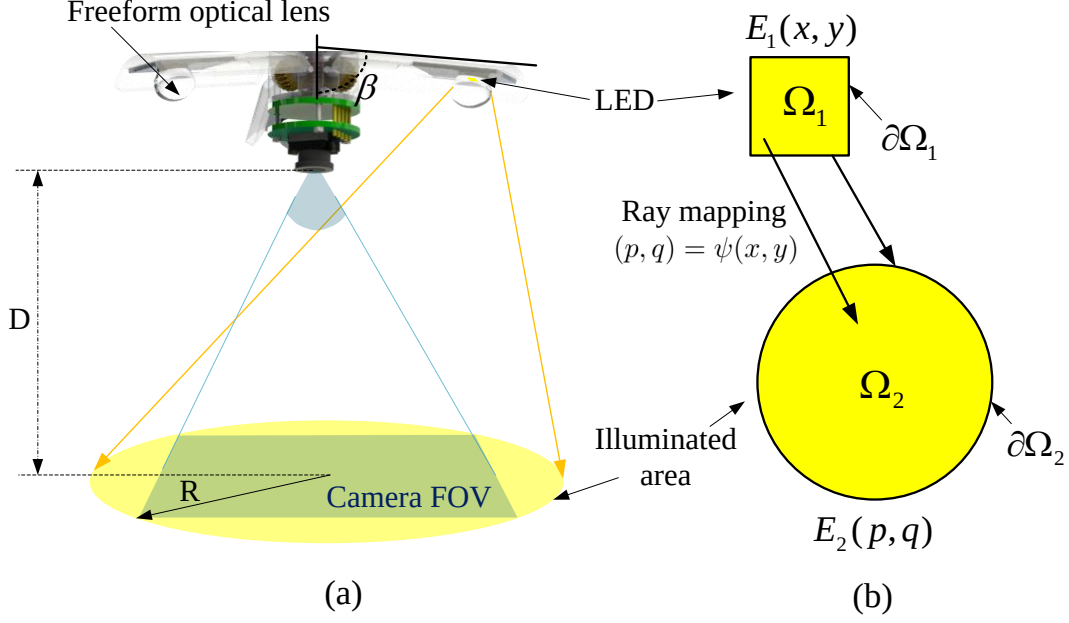


Figure 4.5: (a) Configuration of optimized illumination. The camera’s FOV is contained within a circular illuminated area. (b) Light energy mapping from one LED (domain Ω_1) to the circular illuminated area (domain Ω_2). $E_1(x, y)$ and $E_2(p, q)$ are illuminance distributions in Ω_1 and Ω_2 separately.

flux of 236lm can provide an average illuminance of 11.743lx. But considering that the optical efficiency will not be 100%, using three LEDs will be able to compensate the loss of optical power. Thus, three freeform optical lenses are developed to redirect light beams for achieving the above illumination requirements. The detailed freeform optical lens design technique is proposed in the next section.

4.4 Freeform optical lens design for optimized illumination system

In this section, we present an effective freeform optical lens design method for the LEDs to achieve the optimized illumination requirements. The state-of-the-art solutions to the freeform optical design problem can be categorized into two groups: 1) directly computation; and 2) two-step ray-mapping methods. For directly computation, this problem is governed by a non-standard Monge-Ampère equation[66, 61], which is second-order nonlinear PDE.

Because of the highly nonlinearity of this equation, it is quite challenging to find an easy-to-implement and effective numerical method to obtain a converged solution. Instead of solving the non-standard Monge-Ampère equation directly, two-step ray-mapping methods [86, 23] are studied to establish the relationship between emitted light rays from the light source and incident points on the target plane, and consequently to construct a freeform optical surface. The main challenge of generating a ray map is to enforce integration conditions to guarantee a smooth continuous optical surface. According to the law of conservation of energy, a ray map $(p, q) = \psi(x, y)$ (see Fig. 4.5(b)) is formulated in

$$\int_{\Omega_2} E_2(p, q) dpdq - \int_{\Omega_1} E_1(x, y) dxdy = 0, \quad (4.1)$$

where $E_1(x, y)$ and $E_2(p, q)$ are the LED illuminance distribution in the source domain Ω_1 and the prescribed target illuminance distribution in the target domain Ω_2 respectively. By representing the ray mapping solution as the gradient of a convex potential $\psi = \nabla\omega$, an equivalent form of Eq. (4.1) can be derived as

$$\det \nabla^2 \omega(x, y) = \frac{E_1(x, y)}{E_2(\nabla \omega(x, y))}. \quad (4.2)$$

Numerical solvers to compute Eq. (4.2) are either hard to numerically implement [4, 24], or tricky to get a converged solution [73, 89]. Inspired by the fact that a weak solution of a lower order nonlinear PDE can be approximated by a sequence of higher order quasi-linear PDEs [13], we introduce an effective numerical method to compute the solution of Eq. (4.2) for the freeform optical lens design.

4.4.1 Ray Map Computation

We employ a sequence of fourth-order quasi-linear PDEs, i.e. the biharmonic operator [22], to approximate the numerical solution of the second-order nonlinear PDE in Eq. (4.2). The approximated solution ω^ϵ can be computed from the following quasi-linear PDE with a

Newmann boundary condition (BC):

$$-\epsilon \Delta^2 \omega^\epsilon + \det \nabla^2 \omega^\epsilon - \frac{E_1(x, y)}{E_2(\nabla \omega^\epsilon(x, y))} = 0, \quad (x, y) \in \Omega_1 \quad (4.3)$$

$$\text{BC : } f(\nabla \omega^\epsilon(x, y)) = 0, \quad (x, y) \in \partial\Omega_1 \quad (4.4)$$

where $\epsilon > 0$; f represents the mapping function from $\partial\Omega_1$ to $\partial\Omega_2$; $\partial\Omega_1$ and $\partial\Omega_2$ are the boundary regions of Ω_1 and Ω_2 ; If $\lim_{\epsilon \rightarrow 0^+} \omega^\epsilon$ exists, the weak solution of Eq. (4.3) can be found.

To compute the solution of Eq. (4.3), we initialize a set of ϵ in descent order. The solution of ω^ϵ at the current iteration is initialized by the solution from the previous iteration. By discretizing Eq. (4.3) and (4.4), a system of nonlinear equations is formulated as

$$\mathbf{S}(\mathbf{W}^\epsilon) = 0, \quad (4.5)$$

where $\mathbf{W}^\epsilon$ is a vector that contains the discretized values of ω^ϵ in Ω_1 . The Newton's method is employed to compute the solution of Eq. (4.5).

The illuminated area Ω_2 is set as a circular region with a radius R of 80 mm according to Section 4.3.3. The LEDs are initially modeled as a Lambertian point source $E_1(x, y)$. $E_2(p, q)$ is a function with a constant value in Ω_2 . Optical surface modification is implemented in Section 4.4.2 by considering the extended light source condition.

Fig. 4.6 shows the computed ray maps in different iterations by using 71×71 mesh grids for discretization. To solve Eq. (4.5), an initial guess of $\nabla \omega$ is provided as illustrated in Fig. 4.6(a). It is obvious to see that when $\epsilon = 1$, all the light rays are redirected within $\partial\Omega_2$, as shown in Fig. 4.6(b). But the inner ray-mapping in Ω_2 is not accurate. In the following iterations, the values of ϵ are set as 0.01 and 0.001 to compute more accurate ray mapping results, as shown in Fig. 4.6(c) and (d). The computation is converged at the iteration of $\epsilon = 0.001$.

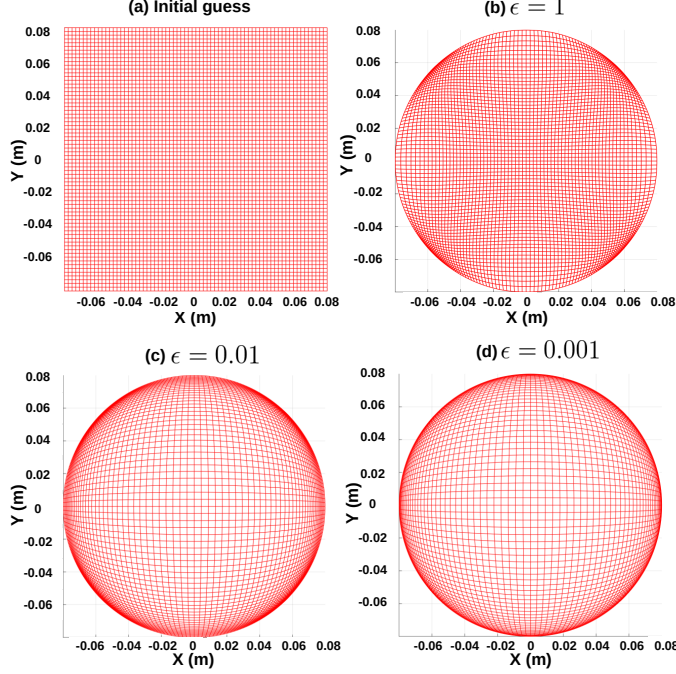


Figure 4.6: Computed ray-mapping results on the target plane by using 71×71 mesh grids. (a) initial guess of the ray map $(p, q) = \nabla\omega(x, y)$; (b) ray-mapping result $\nabla\omega^\epsilon$ when $\epsilon = 1$; (c) ray-mapping result when $\epsilon = 0.01$; and (d) ray-mapping result when $\epsilon = 0.001$.

4.4.2 Optical Surface Construction

Initial surface construction

An initial optical surface is constructed based on the computed ray map shown in Fig. 4.6(d). Each light ray $\mathbf{I}_{i,j}$ in the coordinate system $\Sigma_L\{X_L, Y_L, Z_L\}$ corresponds to a point $T_{i,j}(p_i, q_j, z)$ on the target plane $\mathbf{T}_{i,j}$ in the coordinate system $\Sigma_G\{X_G, Y_G, Z_G\}$, as shown in Fig. 4.7. The relationship between Σ_G and Σ_L is determined by the wings' angle β and the LED positions on the wings. We employ an easy-to-implement method proposed in [86] to construct an initial optical surface. However, due to the accumulated errors, this method can not guarantee the computed normal vectors $\mathbf{N}_{i,j}$ at $\mathbf{p}_{i,j}$ are perpendicular to $\overrightarrow{\mathbf{p}_{i,j}\mathbf{p}_{i+1,j}}$ and $\overrightarrow{\mathbf{p}_{i,j}\mathbf{p}_{i,j+1}}$. We introduce an optimization technique to correct the normal vectors by adjusting the surface points. For each surface point, an optimization function can be constructed as

$$Q_{i,j}(\boldsymbol{\rho}) = \|(\rho_{i+1,j}\mathbf{I}_{i+1,j} - \rho_{i,j}\mathbf{I}_{i,j}) \cdot \mathbf{N}_{i,j}\| + \|(\rho_{i,j+1}\mathbf{I}_{i,j+1} - \rho_{i,j}\mathbf{I}_{i,j}) \cdot \mathbf{N}_{i,j}\| = 0, \quad (4.6)$$

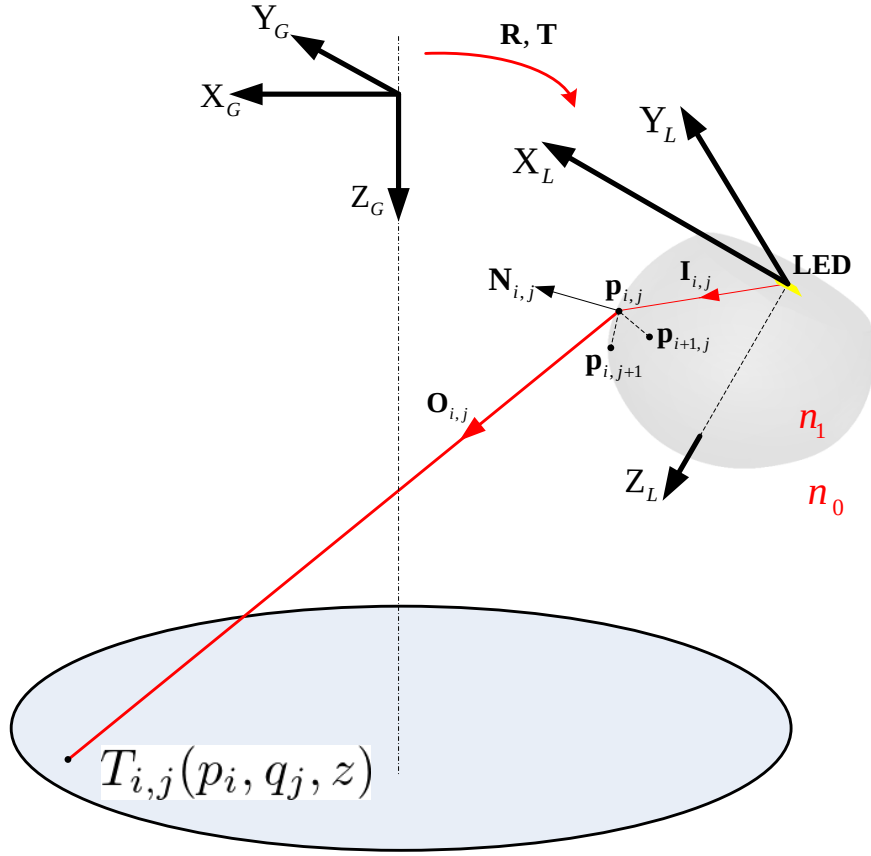


Figure 4.7: Illustration of the initial optical surface construction method. $\mathbf{R}$ and $\mathbf{T}$ represent the rotational matrix and the translational vector from Σ_L to Σ_G .

where $\rho_{i,j}$ represents the distance between the origin of a light source and a surface point. The nonlinear least-squares method is adopted to minimize $\sum_{i=1}^{N_i} \sum_{j=1}^{N_j} Q_{i,j}(\boldsymbol{\rho})^2$. After obtaining the optimized surface points, an optical surface can thus be constructed by using the Non-Uniform Rational Basis Spline (NURBS) [55] method.

Surface modification for extended light sources

The constructed freeform optical surface is actually based on the zero-étendue assumption. In our case, we can not neglect the size of an LED. Based on an initial desired illuminance distribution $E_2(p, q)$ on the target area, a ray map and a freeform optical lens are computed, as shown in the column of "Iteration 0" in Fig. 4.8(a1) and (a2). The resulted illuminance distribution by using the real size LED yields poor uniformity, as shown in Fig. 4.8(a3). The optical performance can be improved by employing a feedback modification scheme [46]. Modified target illuminance distribution is calculated in accordance with the simulated illuminance distributions in previous iterations:

$$\hat{E}_2^{i+1}(p, q) = \frac{E_2(x, y)}{\tilde{E}_2^0(p, q)} \frac{E_2(p, q)}{\tilde{E}_2^1(p, q)} \dots \frac{E_2(p, q)}{\tilde{E}_2^i(p, q)} E_2(p, q), \quad (4.7)$$

where $\tilde{E}_2^i(p, q)$ is the simulated illuminance distribution at the i -th iteration; and $\hat{E}_2^{i+1}(x, y)$ is the modified illuminance distribution for the $(i + 1)$ -th iteration. For more details about our proposed freeform optical lens design method, please refer to [44].

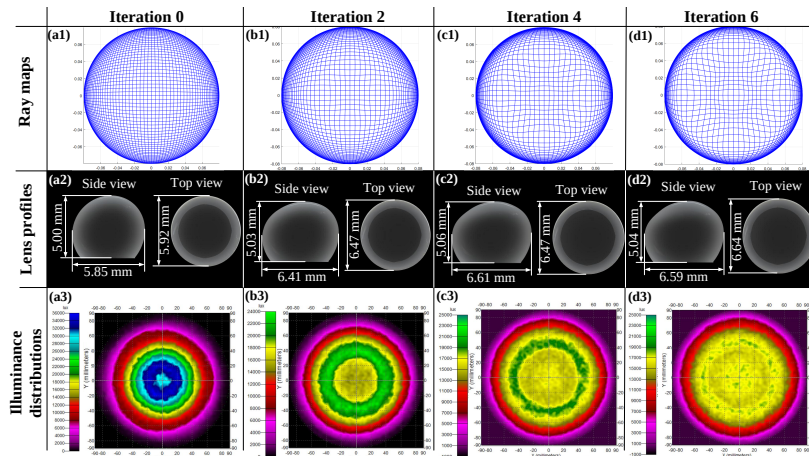


Figure 4.8: Surface modification of freeform optical lenses for the LEDs.

Fig. 4.8 shows the modified ray maps, optical surfaces, and resulted illuminance distributions in different iterations. We sampled the results in iteration 0, 2, 4, and 6 to compare the performance differences. The ray maps and the lens profiles are adjusted based on the simulated $\widehat{E}_2^i(x, y)$ at the current iteration. It is obvious to see that the illuminance uniformity is significantly improved, as shown in Fig. 4.8(a3), (b3), (c3) and (d3). Recall the camera's FOV that is covered by a circular area with a radius R of 57.7 mm when $D=100$ mm in Section 4.3.3. The illuminance uniformity in Fig. 4.8(d3) within the area with a radius R of 60 mm is 94.9%. The average illuminance in this area is 16.132 lx. The optical efficiency is 87.49% considering the Fresnel losses.

4.5 Experiments

4.5.1 Fabrication of the Robotic Camera System

Fig. 4.9 shows the prototype of the robotic camera system including (a) an EACU, (b) an *in-vivo* robotic camera, and (c) a user controller. The specifications of this prototype is presented in Table 4.1.

Table 4.1: Prototype specifications

	Diameter	Height	Weight
EACU	80 mm	50 mm	397.2 g
<i>In-vivo</i> camera	18 mm	72 mm (folded)	43.8 g
		45 mm (extended)	

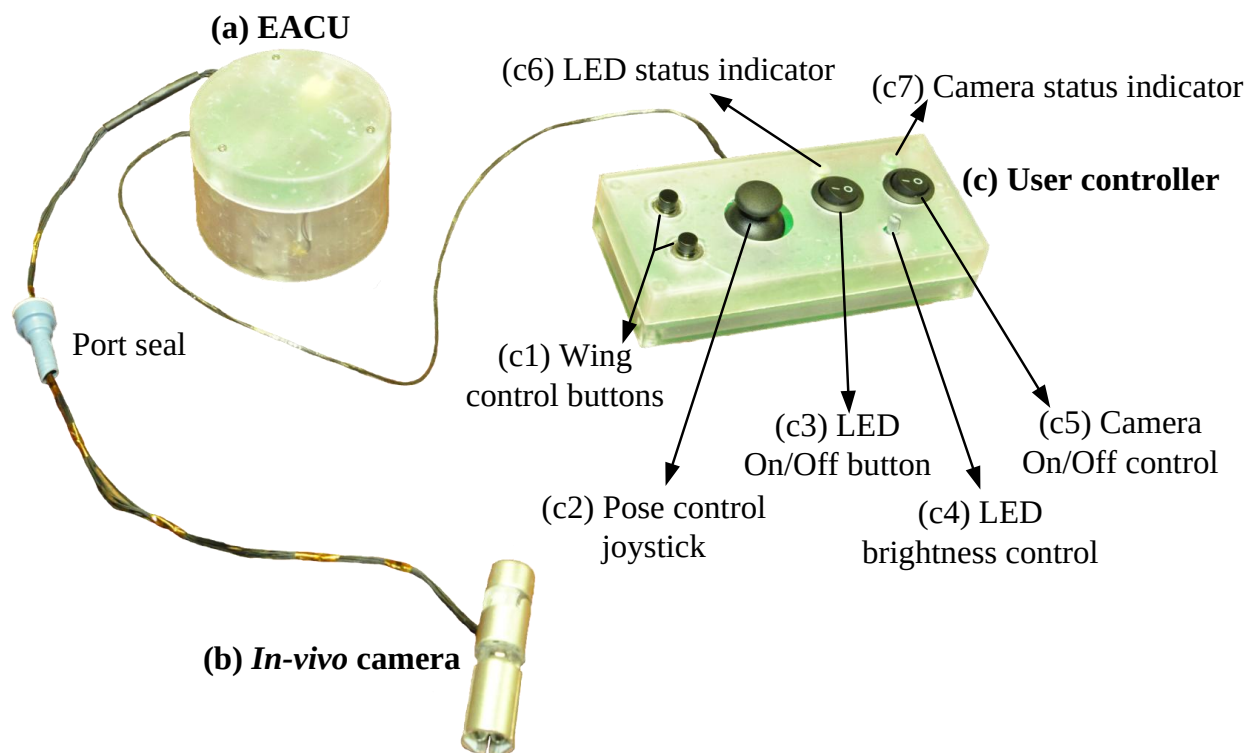


Figure 4.9: Prototypes of (a) the external anchoring/driving unit, (b) the *in-vivo* robotic camera, and (c) the user controller.

4.5.2 Manufacturing of Freeform Optical Lenses

The freeform lens designed in Section 4.4 was manufactured by using an ultra-precision vertical machine (UVM-450C, Toshiba Inc.). The manufacturing accuracy is $\pm 1\mu\text{m}$ which is measured by using both a coordinate measurement machine and an ion beam figuring process. The material of the lenses was selected as PMMA (Polymethyl Methacrylate) with a refractive index of 1.49 and visible light transmittance of 92%. Fig. 4.10 shows the manufactured freeform optical lenses, Cree Xlamp XQ-E white LEDs, and LED circuit boards attached on the wings.

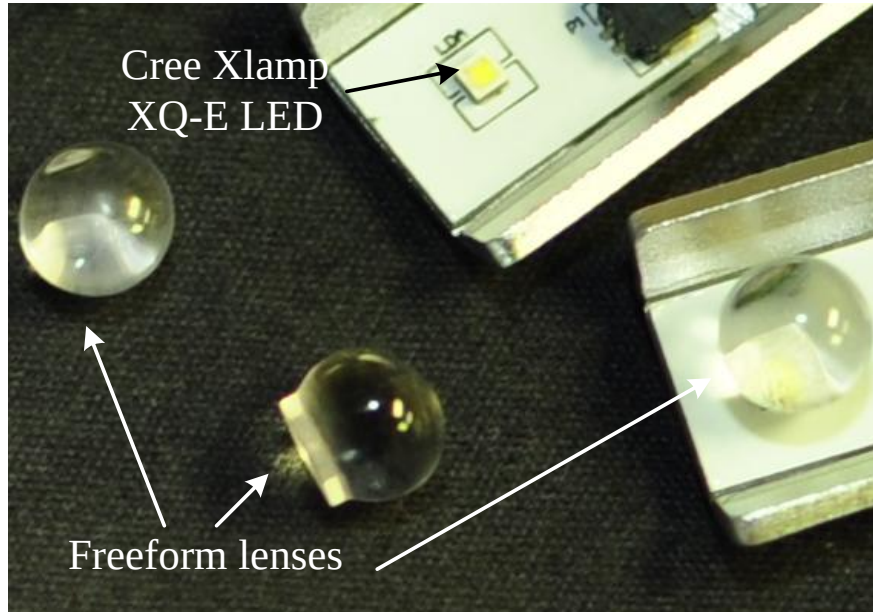


Figure 4.10: Manufactured freeform optical lenses for the LEDs.

4.5.3 Magnetic Coupling Tests between EPM and IPM

The magnetic coupling between the EPM and the IPM provides anchoring, spinning (pan motion), and repositioning functions for the *in-vivo* robotic camera. The EPM and the IPM were customized with the specifications shown in Table 4.2. Considering that the normal range of an abdominal wall thickness is 30 mm~50 mm [70], we need to validate that the anchoring force F_N , spinning torque T_S , and translational force F_x can overcome the camera's gravity F_g , frictional torque T_f , and frictional force F_f respectively, as illustrated in Fig. 4.11.

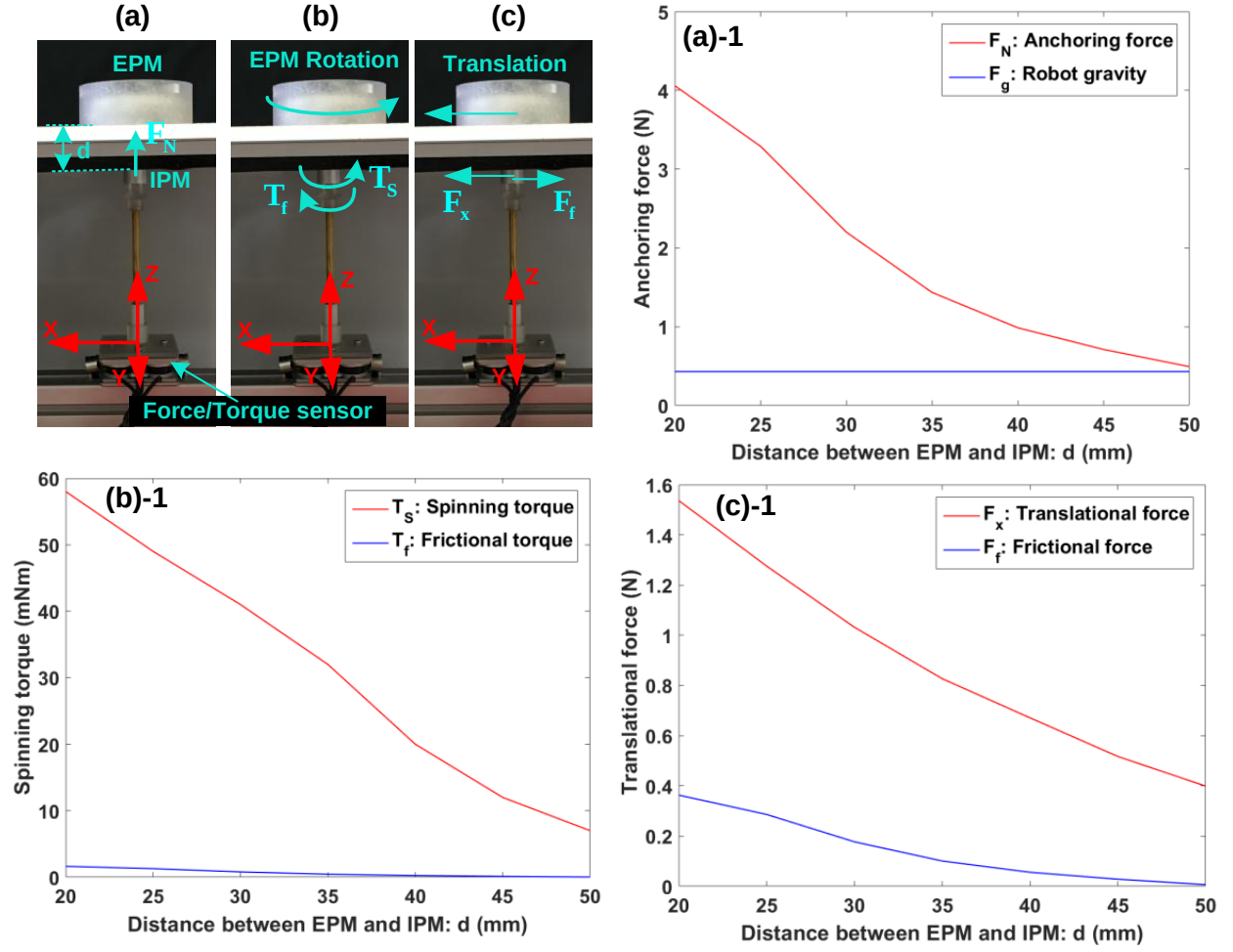


Figure 4.11: Magnetic coupling tests for the EPM and the IPM. (a), (b), and (c) are experiment setups for anchoring test, spinning test, and translational test respectively. (a)-1, (b)-1, and (c)-1 are experiment results of anchoring test, spinning test, and translational test.

Table 4.2: Specifications of the EPM and the IPM

Parameters	EPM	IPM
Material	NdFeB	NdFeB
Grade	N52	N52
Magnetization Direction	Diametrically	Diametrically
Outer diameter	29 mm	16.5 mm
Inner diameter	N/A	2.5 mm
Thickness	30 mm	10 mm

Fig. 4.11(a), (b), and (c) show the experimental setups by employing a six-axis force/torque sensor (HEX-58-RF-2000N, Optoforce Inc.). The IPM was rigidly fixed with the force/torque sensor along Z axis. In the evaluation, we set the distance d between the IPM and the EPM ranging from 20 mm to 50 mm with 5 mm sampling interval.

Fig. 4.11(a)-1 shows the comparison of measured F_N and the robotic camera's gravity $F_g=0.43$ N. The result indicates $F_N=4.06$ N when $d=20$ mm, and $F_N=0.492$ N when $d=50$ mm, which are sufficient to compensate F_g . The maximum pressure applied on the inner abdominal wall 2.31 psi is less than the safe threshold 3.45 psi for preventing undesired histological damage [3]. Fig. 4.11(b)-1 shows the comparison of measured spinning torque T_S and estimated frictional torque T_f . The frictional coefficient between the IPM's case and the simulated abdominal wall is conservatively estimated as 0.1. Comparing to T_S , T_f is ignorable. Fig. 4.11(c)-1 shows the comparison of measured translational force F_x and the frictional force F_f . The EPM was given an offset along X direction to introduce F_x on the IPM. When $d=20$ mm, F_x is measured as 1.53 N, while $F_f=0.36$ N. When $d=50$ mm, F_x is measured as 0.398 N, while $F_f=0.012$ N.

4.5.4 Thermal Test of the Illumination System

Because of employing high-efficiency LEDs in the *in-vivo* robotic camera, a thermal test of the illumination system is necessary to prevent surrounding tissue damages caused by overheating. This prototype contains a thermal management design, which utilizes aluminum PCBs for the LEDs, and the three wings made by anodized aluminum as the heat sink. In the thermal test, we drove the LEDs by using the maximum current 350 mA for 20 minutes

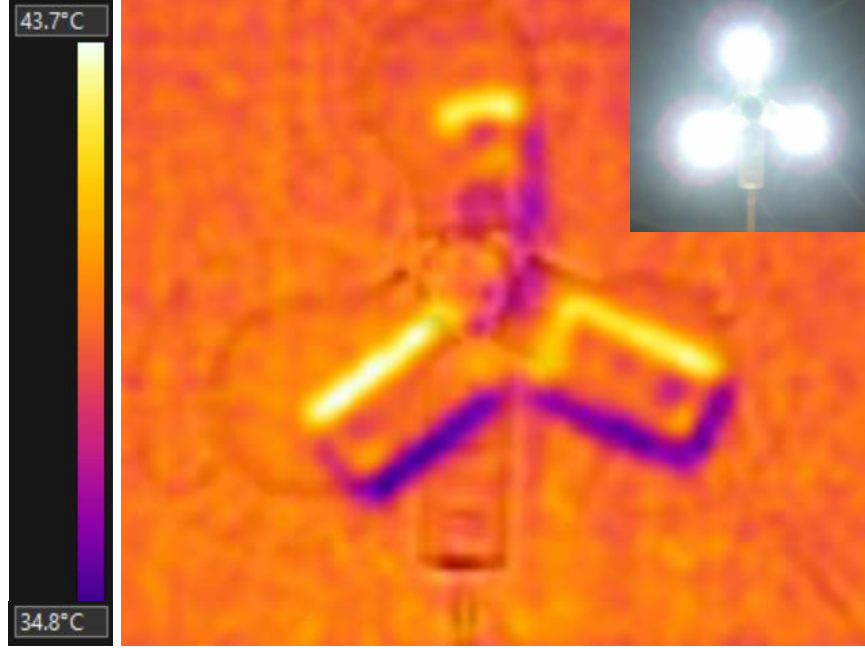


Figure 4.12: Thermal image of the illumination system.

under the room temperature 23 °C. A thermal camera (Duo™ Pro R, FLIR Systems Inc.) is used to measure the temperatures of the illumination system and its surrounding air. Fig. 4.12 shows the measurement result that indicates the maximum temperature 43.7 °C occurred on the wing edges. The surrounding air temperature is about 38 °C. According to [82], when tissue cells are heated to above 45 °C~50 °C, destruction of cells occurs. Therefore, the *in-vivo* robotic camera can be safely manipulated in the abdominal cavity.

4.5.5 Inserting Robotic Camera into Abdominal Cavity

Fig. 4.13 demonstrates the procedure to introduce the *in-vivo* robotic camera into a simulated abdominal cavity via a trocar. After inserting a SILS port (Covidien port, Medtronic Inc.) into the abdominal wall, a customized trocar, which is compatible with the 18 mm outer diameter of our *in-vivo* robotic camera, can be inserted in the SILS port (see Fig. 4.13(a)). Laparoscopic forceps can hold the robotic camera with folded wings to enter the abdominal cavity through the trocar (see Fig. 4.13(b), (c)). The EACU generates magnetic coupling with the IPM, and fixes the robotic camera against the inner side of the abdominal wall. The trocar is removed from the SILS port after fixing the robotic camera in the abdominal

wall. Fig. 4.13(d) shows the working mode of the *in-vivo* robotic camera by extending the wings to expose the illumination system and the imaging system in the abdominal cavity.

4.5.6 Robotic Camera Manipulation Control Tests

Fig. 4.14 shows manipulation control tests of the *in-vivo* robotic camera in the simulated abdominal cavity. Fig. 4.14(a) demonstrates the initial pose of the robotic camera that is anchored against the abdominal wall. To get the robotic camera ready in the working mode, the wings are opened by pushing the wing buttons on the user controller for exposing the camera module and the illumination system in the abdominal cavity, as shown in Fig. 4.14(b). Fig. 4.14(c) demonstrates the tilt motion test of the robotic camera that achieves its maximum angle 49° . Fig. 4.14(d)-(f) show pan motion test of the robotic camera, which rotates along the axis of the IPM. Combining with tilt motion and translational motion, the *in-vivo* robotic camera is capable of visually covering all the surgical areas inside the abdominal cavity.

4.5.7 Performance Evaluation of Optimized Illumination System

In this part, we experimentally evaluate the performance of the optimized illumination system, which includes lighting uniformity, illuminance, and optical efficiency. Fig. 4.15(a) and (b) show the experiment setups for measuring the lighting uniformity and illuminance on the target plane respectively. In Fig. 4.15(a), the robotic camera was attached on a sliding track, which is used for adjusting the distance D between the camera and the target plane. We utilized an acrylic panel with 55% transparency and dimensions of $300\text{ mm} \times 300\text{ mm}$ as the illumination plane. A digital single-lens reflex (DSLR) camera (D90, Nikon Inc.) was set up in front of the illumination plane, and used for recording the illuminance distribution. To measure the illuminance at a specific location, a lux meter (Extech LT-40, FLIR Systems Inc.) was employed. The lux sensor (inside the semi-spherical white cover) was positioned at the height of the camera's axis, as shown in Fig. 4.15(b). A set of measurement points can be accessed by moving the lux meter along the red arrows.

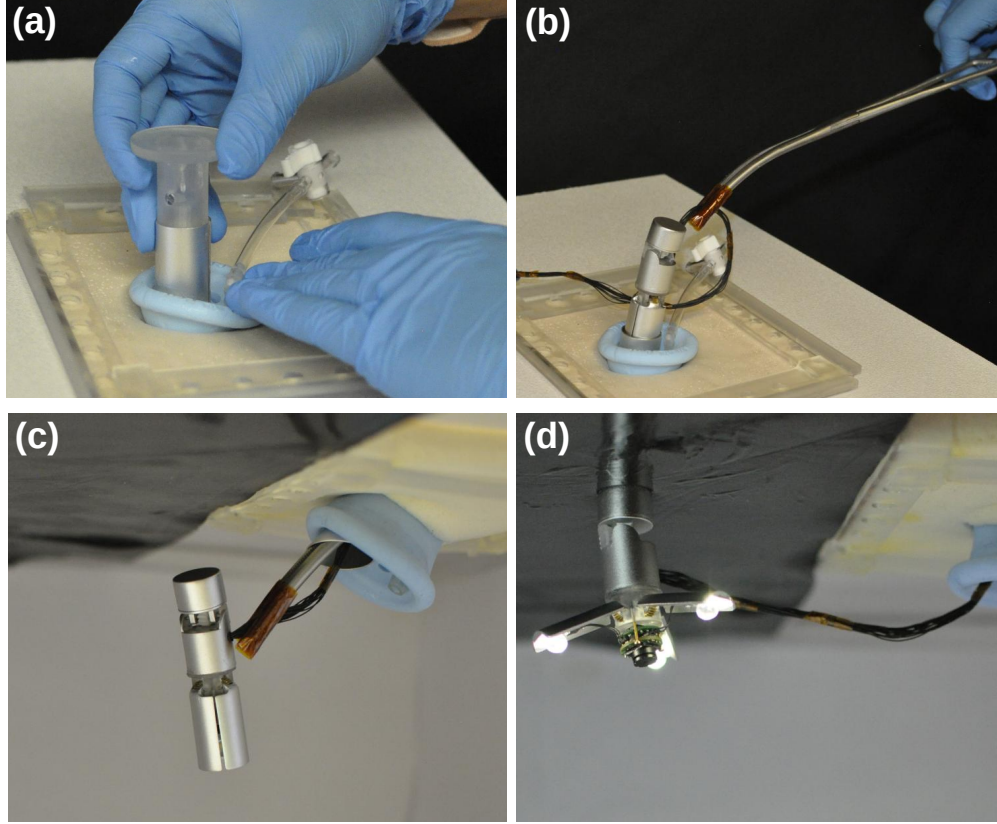


Figure 4.13: Demonstration of the procedure to insert the robotic camera into a simulated abdominal cavity. (a) shows applying a trocar to the SILS port; (b) shows inserting the camera with forceps; (c) shows that the camera is inserted inside the simulated abdominal cavity; (d) shows that the camera is operated in working mode.

Fig. 4.16(a1) shows the captured image of illuminance distribution on the target plane when $D=100$ mm. Considering that the camera's diagonal FOV is 60° , the region of the camera's FOV on the target plane is marked by the red rectangle, which is within the illuminated area. To better visualize the illuminance uniformity, we processed the image in Fig. 4.16(a1) to obtain a contour plot of normalized illuminance distribution, as shown in Fig. 4.16(a2). We sampled the normalized illuminance data across the illumination center (the red dash line in Fig. 4.16(a2)), and generated a 2D plot to show normalized illuminance versus the positions in X direction, as shown in Fig. 4.16(a3). The measured illuminance data as well as the measurement positions are partially indicated in Fig. 4.16(a3). Due to the influence of the extended light sources and the imperfect manufacturing process, the illuminance uniformity is significantly degraded when the illumination radius R is above 60 mm. We have predicted this situation at the beginning of our design. That's why we

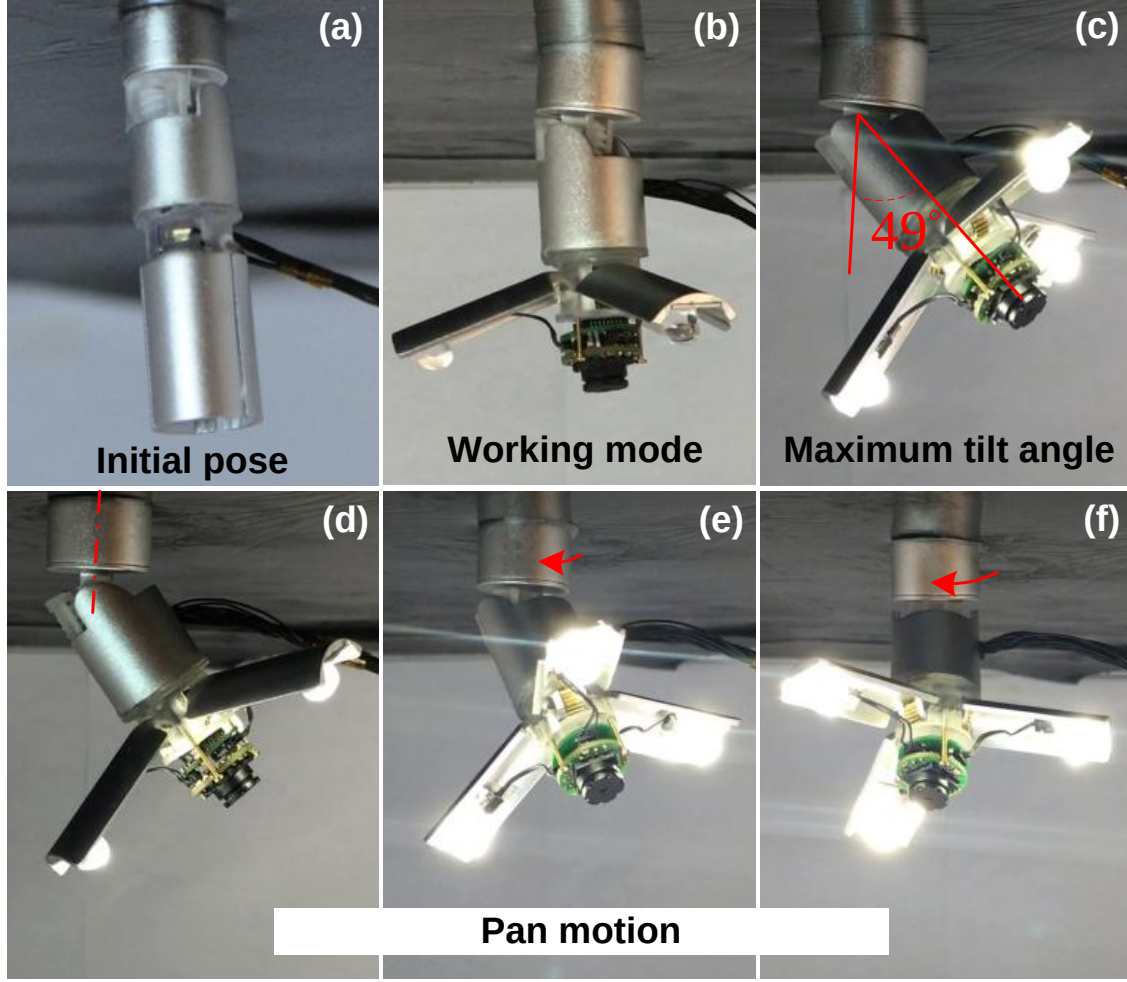


Figure 4.14: Manipulation control tests of the *in-vivo* robotic camera in the simulated abdominal cavity. (a) shows the initial pose of the robotic camera; (b) shows the working mode of the robotic camera; (c) shows the maximum tilt angle of the robotic camera; (d)-(f) show the pan motion control test.

initialized the illumination area with a radius of $R=80$ mm. Similar interpretations apply to Fig. 4.16(b1)-(b3) with $D=80$ mm and Fig. 4.16(c1)-(c3) with $D=60$ mm.

The quantification of illuminance uniformities employs the experiment data in Fig. 4.16(a3) with $x = 90$ mm~210 mm, Fig. 4.16(b3) with $x = 104$ mm~196 mm, and Fig. 4.16(c3) with $x = 115$ mm~185 mm. For each case, the illuminance uniformity U can be computed by

$$U = (1 - \sigma/\mu) \times 100\%, \quad (4.8)$$

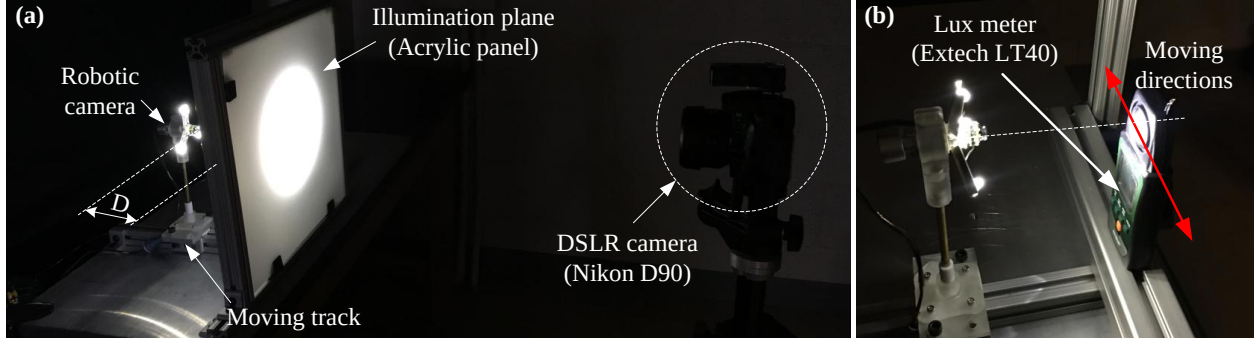


Figure 4.15: Experiment setups for evaluating the performance of the optimized illumination system. The experiment setup in (a) is used for testing the lighting uniformity on the target plane at a distance of D . (b) shows the experiment setup for measuring illuminance values by using a lux meter.

where σ and μ are the standard deviation and the mean value of the sampled illuminance data. To estimate the optical efficiency of the illumination system, we compute the luminous flux projected on the illuminated areas by using

$$\Phi = \int_0^{2\pi} \int_0^{R_i} E'_2(r) \cdot r dr d\theta, \quad (4.9)$$

where $E'_2(r)$ represents an illuminance function modeled by the sampled illuminance data shown in Fig. 4.16(a3), (b3), and (c3). We shifted the y axes to $x=150$ mm for $E'_2(r)$. r denotes the x coordinate in the new coordinate frame. We set R_i as 80 mm, 64 mm, and 48 mm for the cases in Fig. 4.16(a3), (b3), and (c3) respectively. The total emitting luminous flux is 354 lm. An optical efficiency can thus be calculated by $\Phi/354$. In consequence, the performance of the optimized illumination system is quantified in Table 4.3. The illumination system yields illuminance uniformities above 97% within the area of the camera's FOV in all the three cases, which are slightly higher than the simulated result in Section 4.4.2. This situation is caused by limited traced rays (0.5 million) in the optical design software. The average illuminance can reach 14.323 lx when $D=100$ mm, which fulfills the design requirement of a minimum 10.000 lx. While the camera to target distance gets closer, the average illuminance significantly increases. The light brightness control button on the user controller can thus be adjusted for avoiding overexposure. The optical efficiencies are computed as 80.14%, 84.76%, and 85.92% respectively. These values are slightly lower than

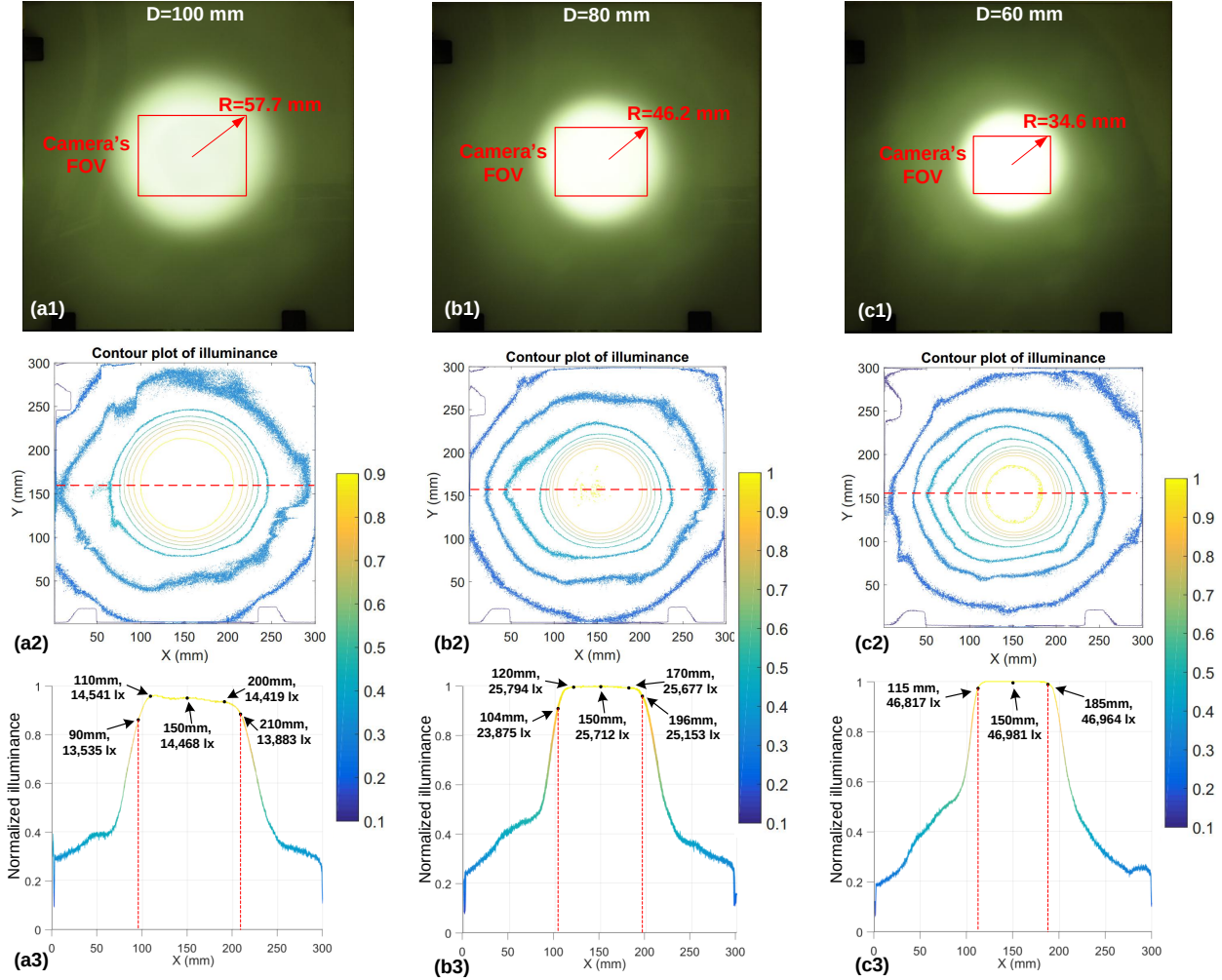


Figure 4.16: Recorded experiment data and analyzed results for evaluating the performance of the optimized illumination system. (a1), (b1), and (c1) show the recorded illuminance distributions on the acrylic plane with the camera-to-plane distances D of 100 mm, 80 mm, and 60 mm. (a2), (b2), and (c2) show the contour plots of the normalized illuminance distributions processed from (a1), (b1), and (c1). (a3), (b3), and (c3) demonstrate the normalized illuminance profiles of the red dash lines in (a2), (b2), and (c2) respectively. Based on the experiment setup in Fig. 4.15(b), the measurement locations and the measured illuminance values are indicated in (a3), (b3) and (c3).

Table 4.3: Performance of the optimized illumination system

	D=100 mm	D=80 mm	D=60 mm
Uniformity	98.17%	97.68%	98.31%
Avg. illuminance	14.323 lx	25.408 lx	46.897 lx
Optical efficiency	80.14%	84.76%	85.92%

the predicted 87.49% in Section 4.4.2. It's most likely because of the manufacturing flaws of the lenses on LED's contacting surface. The optical efficiency is likely to get improved by using injection molding method for future lens manufacturing.

4.5.8 Suturing Task in Simulated Abdominal Cavity

Fig. 4.17 demonstrates the performance evaluation of our *in-vivo* robotic laparoscopic camera for single-port laparoscopic surgery. We carried out a suturing task in the simulated abdominal cavity, which was completely sealed (no extra light sources besides the illumination system). Fig. 4.17(a) shows that the laparoscopic forceps, which handled a curved sewing needle, were guided by the robotic camera to complete the suturing task. Fig. 4.17(b)-(i) are sampled video frames recorded from the robotic camera during the surgical task. As shown in Fig. 4.17(b)-(i), the light rays are uniformly distributed in the camera's FOV with sufficient intensity. This evaluation provides an initial evidence to demonstrate the effectiveness of our *in-vivo* robotic laparoscopic camera.

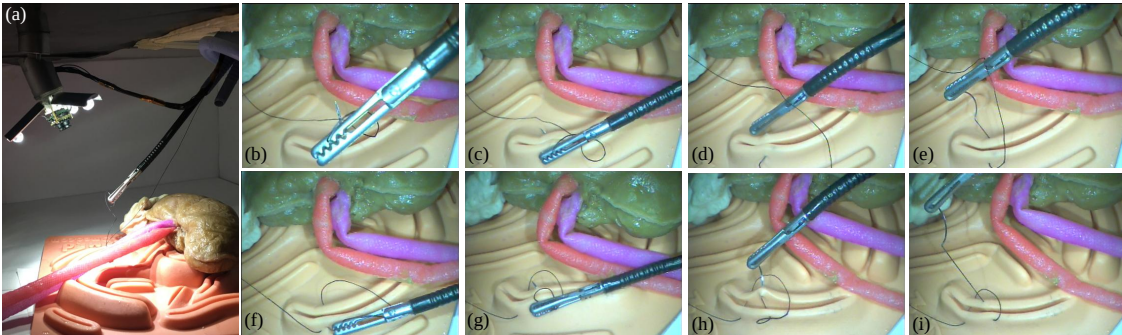


Figure 4.17: Performance evaluation of the robotic camera by implementing a suturing task in the simulated abdominal cavity. (a) shows the instruments setup of the surgical task; (b)-(i) demonstrate the sampled images recorded by the robotic camera.

4.6 Summary

In this chapter, we present an initial prototype of a transformable *in-vivo* robotic camera system integrated with an optimized illumination system for single-port laparoscopic surgery. This prototype includes an EACU, an *in-vivo* robotic camera, and a user controller. The key innovation of this work is to introduce freeform optical lenses into the *in-vivo* robotic camera for controlling light beams from the LEDs to achieve optimized illumination. To design freeform lenses, we propose an effective ray-mapping based numerical computation method for the freeform optical lens design. The resulted illumination system features greater than 97% illuminance uniformity, greater than 80% optical efficiency, and greater than 14.323lx illuminance on the target area when the camera-to-target distance ranges from 50 mm to 100 mm. In the experiment, we demonstrate controllability of the robotic camera, performance of the optimized illumination system and effectiveness to apply the robotic camera in a suturing task by using a simulated abdominal cavity.

Further improvements based on this prototype will include waterproof design of the *in-vivo* electronic system, reducing tethered wires for more flexible manipulation, sterilization compatible design, etc. Considering the limitation of the compatible abdominal wall thickness range (i.e. 20 mm~50 mm) in this initial prototype, a series of EPM models in the EACU will be developed to cover the wide range of patients' abdominal wall thicknesses.

As what we discussed in the introduction, the ultimate goal of this research is to develop a high performance *in-vivo* robotic camera with superior imaging quality to compete with state-of-the-art “long-stick” laparoscopes. The achieved optimized illumination is only one step towards this goal. We are currently developing an optimized imaging system combined with the developed illumination system to achieve superior laparoscopic imaging quality.

Chapter 5

Clear Imaging for In-vivo Camera Systems

5.1 Abstract

Robotic systems have recently drawn attention in minimally invasive surgeries due to their increased dexterity feature. A major drawback of these systems is image blurring due to lens contamination which cause imaging impairment during up to 40% of surgery time. This work demonstrates a novel laparoscopic magnetic driven camera system with implemented *in-vivo* lens cleaning and debris prevention systems. This camera robot can cover 150 degrees field of view inside the abdominal cavity and provide adjustable illumination system to improve the video quality. Design details for different modules, such as anchoring, actuation, video capturing, illumination, debris prevention and lens cleaning of this robot have been provided and discussed. This camera robot can decrease the possibility of lens contamination by creating CO₂ gas barrier in front of lens. In case of contamination it can clean the lens *in-vivo* without removing the camera from abdominal cavity. The lens cleaning module has been tested for water vapor and water droplets. The robot is manufactured and each module has been validated by designed experiments.

5.2 Introduction

Minimally invasive surgery (MIS), single incision laparoscopic surgery (SILS) and natural orifice transluminal endoscopic surgery (NOTES) are becoming more popular as replacements for traditional open surgeries. These methods benefit patients with lowering blood loss and post-operative pain, reducing recovery period and hospital stay time, decreasing surgical area scarring and cosmetic issues, and lessening the treatment costs, hence greater patient satisfaction would be earned [32, 37, 72]. In all aforementioned methods, the surgeons control and manipulate the surgical instruments, including gripping, cutting and suturing tools, through constrained access port/ports. The ports are created by surgical incisions over patient's anatomy, usually periumbilical, i.e. which pierces the umbilicus or navel, ranging from 5 mm to 12 mm in MIS [48] and 15 mm to 35 mm in SILS [37]. Manipulating surgical instruments from outside of surgical spot and performing surgery needs precise hand-eye coordination which is provided by insertable cameras. These cameras are usually inserted

by a standard trocar through one of the incisions, mostly less than 25 mm in diameter, into the abdominal cavity to provide the video feedback for the surgeon to operate. Recently, *In Vivo* robotic camera systems are developed that can be fully inserted into the abdominal cavity and provide real-time visual feedback for surgeons.

A noteworthy drawback of *In Vivo* robotic camera systems is clear visualization during the MIS operations. The inserted camera lens is vulnerable to get dirty often with intra-abdominal fluids, blood, surgical rinsing fluid, operation dust particles and smoke which results in image impairment [8]. Researches have shown that approximately one third of the minimally invasive surgeries are performed without clear visualization for surgeons [92]. However, to the best knowledge of authors, None of the robotic systems have considered a prohibitive and/or cleaning mechanism in case of lens contamination. Image impairment reduces the hand-eye coordination and surgeon's dexterity as a result. Also it causes the interruption in the surgery work flow which influences the performance of surgical team. In the following we will elaborate the detailed state of the art for laparoscopic lens cleaning systems.

The standard procedure of lens cleaning during a laparoscopic surgery is: 1) removing the camera from abdominal cavity through port; 2) wiping the lens with gauze; 3) defogging the lens with hot water or solution. These steps can cause a delay, interrupt the surgery work-flow, increase surgery time and increase the costs as a result. Unnecessary interruptions can cause anxiety and errors in judgment, technique and may potentially cause injury to the patient. Lens removal and cleaning is the most frequent cause of interruption in advanced surgeries [8] which results in longer operation time and increased costs.

Condensation on the lens frequently happens due to differences between room and intra-abdominal temperatures. Despite using anti-condensation solutions, it can not be solved entirely [85]. Removal of lens will cause a secondary issue and that is the port contamination, which requires extra time for cleaning per each lens removal [77]. In rare cases, some patients are allergic to povidone-iodine solution used for lens defogging when it contacts the Intra-abdominal [62].

Some researchers have addressed these issues by providing *In Vivo* cleaning for conventional laparoscopic surgeries. EndoClear is a device that can be attached to the

internal abdominal wall in the beginning of surgery and be used as a cleaning station during surgery [8]. Disadvantages of this system are: 1) it needs fixation inside the cavity which probably done by suturing which is harmful; 2) the EndoClear itself can get dirty during surgery due to the probable bleeding and fluids inside the abdominal cavity.

An Irrigation system has been developed to wash the lens during surgery [47]. The drawback of irrigation system is that it needs a fluid tube to be inserted along the trocar or an overtube which influences the single port dexterity by increasing port conflict, and may eventually cause larger incision sizes on patients body.

Floshield is a product which uses a plastic overtube that fits over conventional laparoscopes and blows CO₂ and fluid solution over the lens of the laparoscope [7]. Floshield itself increases the trocar outer diameter through the entry port which reduces the dexterity, also it needs external fluid pumping through tubes.

A mechanical wiper has been implemented to clear the lens for *In Vivo* visual robot system with digital defogging by Feng et al [21]. The drawback of this design is the complexity of mechanisms, transferred from motor to actuate the wiper. Additionally, in this system the camera lens is easily exposed to any contamination source and no prevention system has been considered for this purpose.

Digital defogging system has also been used to improve the imaging quality [30]. Although this is an interesting approach, it won't be effective in other contamination types.

This chapter proposes a novel, *In Vivo* gas generated debris prevention system to provide continuous clear imaging during MIS and SILS, without interrupting the surgery flow for robotic camera systems. The twofold functions of the proposed cleaning system are: Firstly, it deviates the approaching impairment sources toward the camera lens; Secondly, if the image is impaired by contamination, a mechanical system would be actuated to retract the LED wings, which guides the generated gas to ward the camera lens through a designed edge on a wing. This stream helps to wipe the image impairment cause off the lens surface. This lens cleaning system is designed and mounted on a modular *In Vivo* laparoscopy robotic camera, which is presented in details in section 5.3.

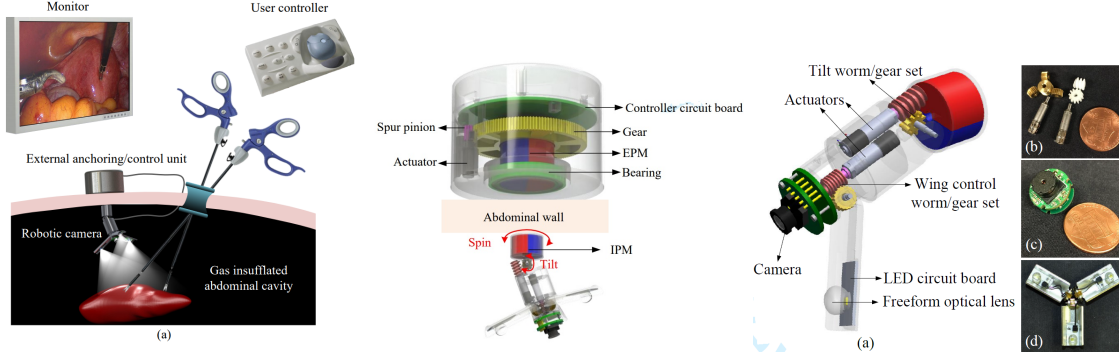


Figure 5.1: Camera Robot Concept and Design

5.3 Modular, *In Vivo* Laparoscopy Camera Robot

In this section we briefly discuss the design of a laparoscopic robotic camera systems, on which the cleaning system is mounted. This robotic system is used for demonstrating the *In Vivo* function and experimental study of the cleaning system.

5.3.1 Robotic Camera System Design

A major challenge in SILS and NOTES is the shared use of incision for both surgical instruments and camera trocar. Since all tools enter the operation area through a shared port, they are almost parallel with each other which reduces degrees of freedom of each tool. This will make the control of surgical tools more complex and as a result, dexterity will be degraded. As a result, the ease of operation for surgeons is impaired by loss of triangulation and increase of clashing between surgical devices at the shared entry port.[16, 28, 75]. Besides that, the quality of video feedback might be decreased, since the camera's triangulation may not cover the proper view angles needed for the surgery by traditional trocar pushing. Hence, these constraints may cause muscular, Mental and eyesight fatigue for operating staff.

To address these issues, A camera robot system is designed to move inside the abdominal cavity by magnetic manipulation, which consists of six modules as:

1) anchoring module: enables fixation of the robot inside the cavity with permanent magnets and provides motion upon interior side of abdominal wall with magnetic fixation. Pan rotation of the robot is also provided by magnetic coupling;

2) actuation module: consists two actuation mechanisms for tilt motion and wing spreading. Tilting angle of 0 to 45 degrees helps the camera to cover the whole abdominal cavity area. Spreading mechanism controls the angle of wings for proper illumination and simultaneously used for actuation of wiping mechanism;

3) video capturing module: consists of a CMOS imaging sensor (OmniVision OV7850) [CMO] and a low-profile pin-hole lens (Sunex PN DSL871) with 56° nominal field of view;

4) illumination module: consists of a LED circuit board and a reflector attached on each wing. The reflectors are used to improve the optical efficiency. The three LED wings can be adjusted to the desired angle by the wing spread mechanism to provide proper illumination requirements;

5) dirt and debris prevention module: consists of a micro-pump which inlets the CO₂ gas from the insufflated abdominal cavity and blow it through the designed channel toward the camera lens;

6) Debris removal module, as shown in Fig.5.1(b). Total length of the camera robot is 90 mm in retracted mode and 60 mm in spread mode. Considering the tilt ability of camera robot and wing spread, it can be fitted in the 100 mm to 150 mm in depth workspace provided inside the abdominal cavity after insufflation. In this section design details of each module will be discussed.

5.4 Clear Imaging and *In Vivo* Contamination Challenge

Visualization clarity is a major issue in MIS which influence the surgery performance as thoroughly discussed earlier. In this section, the proposed solutions for maintaining clear image during robotic laparoscopic operation will be discussed in details. The main idea is to deviate the possible contaminations and particles from reaching the camera lens and preventing the image impairment. Based on this idea, An *In Vivo* debris prevention system is designed, modeled and studied for different contaminating situations. A micro piezo-electric pump generates a gas barrier in front of camera lens through an optimized nozzle chamber.

This system actively blocks the approaching obstacles toward the lens (such as: blood, water, smoke and etc.), while the camera provides surgical area's video feedback to the surgeon.

The novelty of this design is providing a continuous prohibitive gas barrier in front of the lens, while this flow is generated totally *In Vivo* and there is no need for increasing the incision size or adding any extra tubes, which increases the instruments outer diameter, compared to other mentioned methods. Additionally this system resolves the ex-vivo gas generation systems, manually operated or electric pumps, and decreases the possible ex-vivo complexity for surgery team. This novel *In Vivo* debris prevention module can decrease the surgery time by decreasing the required lens cleaning during MIS.

5.4.1 *In Vivo* Contamination Sources

A proper design of debris prevention system requires the knowledge of probable lens impairment sources during MIS, inside the abdominal cavity. During laparoscopic surgeries, contamination sources can be categorized as: 1) fluids: blood, peritoneum fluid, surgical washing water and saline; 2) gas: surgical smoke caused by cauterization and bone shaving dust; 3) Fogging: caused by water vapor and mainly temperature difference of surgery room and abdominal cavity, 16.4°C and 36.8°C respectively.

Fogging deteriorate the vision by condensation on camera lens surface, while surgical smoke and dust mostly block the camera's field of view. The fluids contamination may happen due to lens contact to interior parts or fluid splash over the lens. The prevention system is designed to effectively deviate the approaching fluid or gas streams, while simultaneously can remove the lens fogging with the gas circulation around the lens. The debris removal system removes the accumulated particles from lens surface which we discuss later. Here we provide a brief overview of the aforementioned sources characteristics.

Surgical smoke, also known as cautery smoke or electrosurgery smoke, is produced by interaction of mechanical and heatproducing instruments with tissue, such as dissection and hemostasis tools. This smoke is gaseous byproducts of the disruption and vaporization of tissue protein and fat [53]. A research performed by Nicola et al. shows that the surgical smoke rises with the speed of 9 m/s to 18 m/s. The kinetic energy in smoke particles makes them able to reach up to the distance of 0.87 m from burning spot [52].

The 4.7 to 5.5 liter of blood an adult continually pumps, flow at an average speed of 1.3 m/s to 1.8 m/s in veins. Blood velocity is different in veins which is a factor of physiological and geometrical parameters. The MIS surgical site, mainly abdominal cavity, contains different types of veins and blood velocities as well. Table 5.1 presents different vessel types passing through abdomen and blood flow mean velocity. Any vessel rupture during surgery may cause a blood spray inside the cavity which might result in vision impairment.

Table 5.1: Blood Velocity in arteries and veins

Vessel	Mean Velocity (m/s)
Abdominal aorta	0.08 - 0.20
Vena cavae inferior and superior	0.05 - 0.1
Capillaries	0.0003

Considering the gaseous and liquid dirt sources characteristics, the designed prevention system should be able to effectively divert them during the minimally invasive surgeries. In the following the modeling and design details of this system are presented.

5.5 Debris Prevention System

Diverting the approaching contamination sources toward the imaging module, requires an active repelling mechanism. The proposed solution in this chapter is continuous blowing of insufflation gas with an embedded pump in the camera robot system. The insufflation gas, c , was selected since it makes the system needless of any *ex vivo* material source. Additionally, continuous CO_2 stream would not impair the image during surgery, unlike the other fluid options like water. The proposed system uses a micro pump, which blows the gas through a chamber and generates a gas barrier in front of lens through nozzle. This section discusses

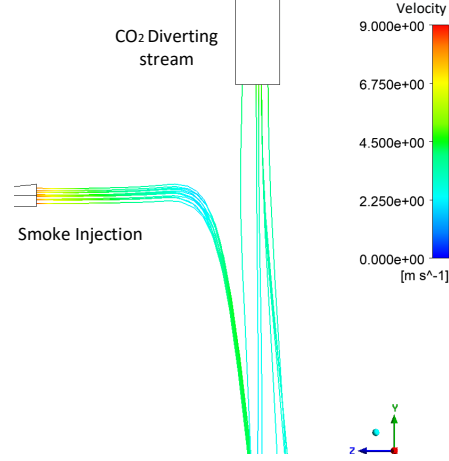


Figure 5.2: Simulation study in Ansys CFX to figure out the minimum required diverting stream velocity.

the mathematical modeling of the gas barrier and mechanical design of the debris prevention system.

5.5.1 Diverting Gas Barrier Requirements

Knowing the contaminating sources characteristics, the first step would be finding the required flow properties of diverting gas barrier which would be able to repel them. A preliminary simulation study was conducted to find out the minimum flow speed required for diverting the smoke, water and blood streams discussed in section 5.4. As shown in Figure 5.2, a CO₂ insufflated abdominal cavity is simulated in ANSYS CFX, with the preset pressure of 1.6 kPa and temperature of 37 Celsius [14]. Simulated smoke, water and blood streams, with different angles are injected toward the CO₂ stream, blown from a linear nozzle with 1 mm exit gap. Table 5.2 shows the minimum required CO₂ stream velocity for diverting the approaching streams. We conclude that the diverting gas barrier requires a minimum stream velocity of 3 m/s in different contaminating scenarios.

5.5.2 Gas Barrier Geometry

The main purpose of the Prevention system is to generate continuous gas stream in front of camera lens to prevent lens contamination, in order to avoid any cleaning required during operation. Although the camera's field of view is only 56°, as shown in Fig. 5.3, the

Table 5.2: Minimum stream velocity required for diverting

Contamination source	Min. Stream Velocity (m/s)
Smoke (Max. 9 m/s)	3
Water (Max. 0.5 m/s)	2.3
Blood (Max. 0.2 m/s)	0.9

prevention area is a hemisphere centered by camera lens. Any type of dirt approaching the camera lens through this space must be deflected to maintain the clear vision. In order to generate this shape of gas barrier, the best solution would be using an annular nozzle to cover the whole area around the camera lens homogeneously. An annular nozzle is well aligned with the circular design of the camera lens, as well as the circular design of robotic camera. Dimensional constraints of the chamber and nozzles are: 1) Minimum inner diameter of the nozzles is 9 mm, due to outer diameter of imaging board; and 2) Maximum outer diameter of chamber could be 10.5 mm, due to inner diameter of LED wings. Preliminary simulation results of commercial ANSYS 18.2 presented in Fig. 5.3(b) shows the homogeneous nozzle outlet from an annular nozzle, which will be used for gas barrier design.

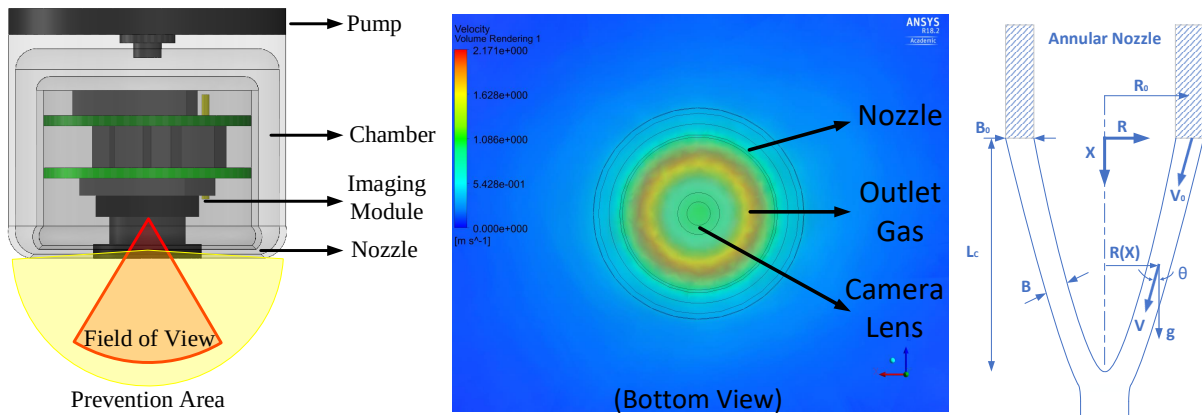


Figure 5.3: Debris Prevention System Concept design - Annular nozzle outlet profile - Annular jet profile parameters

5.5.3 Gas Barrier Mathematical Modeling

The outlet of an annular nozzle will shape an annular jet, with radius R , which varies by nozzle radius R_0 , nozzle outlet gap B_0 , outlet gas velocity V_0 , gravity g , surface tension σ and pressure difference on the two sides of the jet ΔP .

$$R = f(x, R_0, B_0, V_0, g, \rho, \sigma, \Delta P) \quad (5.1)$$

In a thin walled annular nozzle, as shown in Fig. 5.3(c), if the outlet gas thickness is small enough compared to nozzle diameter, $R_0/B_0 \gg 1$, the governing fluid dynamic equations in cylindrical coordinates [33] are:

1) Continuity equation:

$$BRV \cos \theta = B_0 R_0 V_0 \cos \theta_0 \quad (5.2)$$

where V is the local gas velocity with the angle of θ and thickness of B . R refers to annular flow radius in the radial coordinate.

2) Axial momentum equation:

$$\ddot{X} = g - \frac{2\sigma}{\rho_{co_2} BR} \sin \theta \cos \theta - \frac{2\sigma}{\rho_{co_2} BR_c} \quad (5.3)$$

where X is the vertical coordinate and R_c is the radius of curvature of the flow in the vertical plane.

3) Radial momentum equation:

$$\ddot{R} = \frac{2\sigma}{\rho_{co_2} BR} \cos^2 \theta - \frac{2\sigma}{\rho_{co_2} BR_c} \cos \theta \quad (5.4)$$

In our debris prevention nozzle design, since we have a near-vertical annular flow, the angle of θ is always very small, except near the closure area. The inner and outer pressure of the flow is assumed equal. Additionally, the flow radius, R , and the radius of curvature, R_c , are assumed to be positive. Considering these assumptions and non-dimensionalizing, the equations 5.1-5.3 become:

$$brv = \frac{V_0^2/(gR_0)}{(R_0/B_0)} = \frac{Fr}{K} \quad (5.5)$$

$$\frac{d^2x}{dt^2} = V_0^2/(gR_0) = Fr \quad (5.6)$$

$$\left[1 - \frac{r}{We(1+t)}\right] \frac{d^2r}{dt^2} + \left[\frac{r}{We(1+t)^2}\right] \frac{dr}{dt} = -\frac{Fr^2}{We}(1+t) \quad (5.7)$$

where Fr is Froude number and We is Weber number, $We = \rho_{\text{co}_2} B_0 V_0^2 / 2\sigma$. All lengths are made dimensionless with dividing by R_0 , time by V_0/g and velocities by R_0g/V .

By finding the dynamic equations of the fluid flow, we can find the outlet flow profile, $r(t)$, and closure length, L_{cl} . $r = r(t)$ can be calculated by numerical integration of equation 5.6. Closure happens when the annular flow radius at closure point, r_{cl} , equals half the thickness of flow at that point, $r_{cl} = b_{cl}/2$. Since the equation 5.5 is integratable, we can find the vertical coordinate as a function of Froude number and dimensionless time:

$$x = Frt\left(1 + \frac{t}{2}\right) \quad (5.8)$$

The closure point radius can be found by solving equations 5.5 and 5.8 as:

$$r_{cl} = \sqrt{2K(1+t_{cl})} \quad (5.9)$$

where t_{cl} is the closure time and can be found by solving 5.9 and numerical solution of 5.7. Closure length, L_{cl} , can be calculated from 5.8 as:

$$L_{cl} = Frt_{cl}\left(1 + \frac{t_{cl}}{2}\right) \quad (5.10)$$

Equation 5.10 shows that the closure length is a function of Froude number and can be controlled by changing the outlet gas velocity V_0 , and nozzle radius R_0 .

The nozzle radius is considered fixed in this study based on the camera dimension and the robot outer diameter. The closure length is calculated based on different outlet gas

velocities both numerically and in Ansys CFX simulation. Graph 5.4 depicts the numerical calculations of the closure lengths for different outlet velocities V_0 .

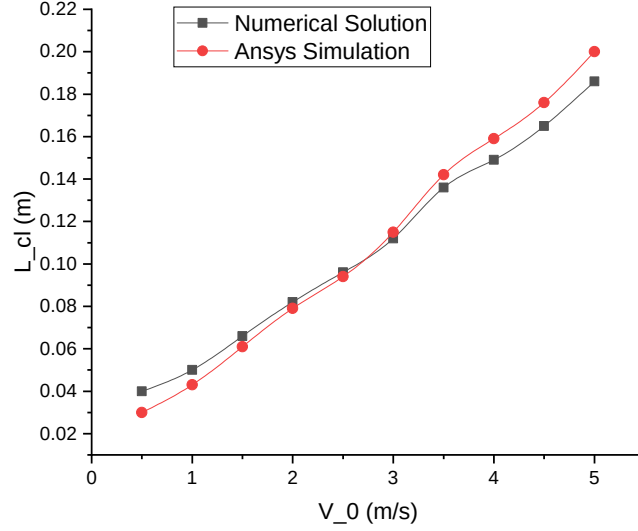


Figure 5.4: Numerical calculation and simulational study of closure length based on outlet velocity V_0 .

5.5.4 Pump Selection and Control

In this chapter, we are using a piezo-electric micro pump for generating airflow inside the abdominal cavity, (hereafter we call it simply "pump"). This novel pump features: 1) a compact size of 20 mm in diameter, with less than $2\mu\text{m}$ deflection of diaphragm center, which makes it suitable for minimally invasive surgeries, (within the range of standard incisions, $<30\text{ mm.}$); 2) high performance in supply flow rate with respect to static pressure, regardless of pressure drop in chamber's narrow channels; 3) the piezo diaphragm of this pump operates at its third vibration mode with the frequency range of $24\text{ kHz} < f < 27\text{ kHz}$ which is beyond the human hearing range and will help acoustics during the surgery; 4) experimental vibration Analysis of the pump shows no defects on image quality during actuation of debris prevention system, which makes it ideal for this surgical camera system.

The designed pump uses a ceramic piezo-electric disk with 11 mm in diameter, attached to a brass plate. While the pump activates, the diaphragm starts to operate at its third

normal mode of vibration. Pump's first normal mode of vibration is amongst hearing range of human and was omitted to help surgical area acoustics. Fig.5.5 shows conceptual mechanism of the pump, as the gas enters the operation area in front of diaphragm through the narrow side channels. The piezo diaphragm is activated with square wave signal with driving voltage ranging from 10 v to 20 v. The piezo's resonant frequency is unique for each sample and must be determined to find the maximum flowrate outlet of the pump. Outlet velocity is measured with a hot wire anemometer placed in the distance of 0.5 cm from nozzle to show the immediate velocity outlet of pump. The Anemometer uses micro-glass bead thermistor with resolution of 0.01 m/s and measurement range of 0.20 m/s to 40.00 m/s. These measurements are conducted for different pumps around their nominal resonant frequency of 25 kHz. Fig.5.6 shows the measured outlet flowrate of three tested pumps to find the maximum outlet frequency.



Figure 5.5: Micro piezo-electric pump (a) Micro pump actuation concept. (b) Pump design and dimensions. (c) Pump analysis setup: Anemometer used for measuring the immediate velocity of pump outlet.

The piezo-electric pump's outlet flowrate is a variant of driving voltage. The nominal driving voltage (peak to peak) of square wave signal for piezo layer is in the range of 10 v to 20 v. Fig.5.5 depicts the driver circuit to generate the driving voltage signal for the pump. The pump analysis setup shown in Fig.5.5 was used to study pump's outlet flowrate with respect to driving voltage. It should be mentioned that in this test, for each pump, the frequency was set at its resonant frequency as measured in Fig. 5.6. This study shows that increasing the driving voltage increases the flowrate outlet of pump as illustrated in Fig.5.7.

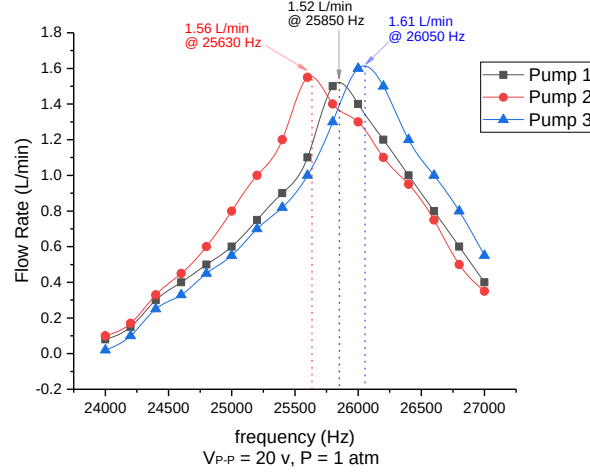


Figure 5.6: Outlet flowrate of the pumps with respect to driving frequency with driving voltage of 20 v and pressure of 1 atm.

5.6 Prototype Development and Experimental Investigation

In this section, we will discuss the performance of manufactured camera robot and proposed cleaning and prevention systems. Experiments have been designed to replicate the *in-vivo* dirt sources to prove the proposed concepts.

5.6.1 Camera Robot Manufacturing

The robot prototype was manufactured by 3D-printing, with the resolution of 0.025 mm. Two brushless DC motors were implemented to actuate the tilt mechanism, spread mechanism and cleaning mechanism as well. Two worm-gear sets were designed to enhance the generated torque. The spread mechanism uses one worm to spread three wings simultaneously. The Fig.5.8(a) shows manufactured modules and assembled mechanisms.

5.6.2 Magnetic Coupling

In this section the magnetic coupling of EPM and IPM is tested. The total camera robot with all the modules weigh 58.320g. The EPM is able to anchor the camera robot from the distance more than 46mm which proves the ability of using this system over the abdominal

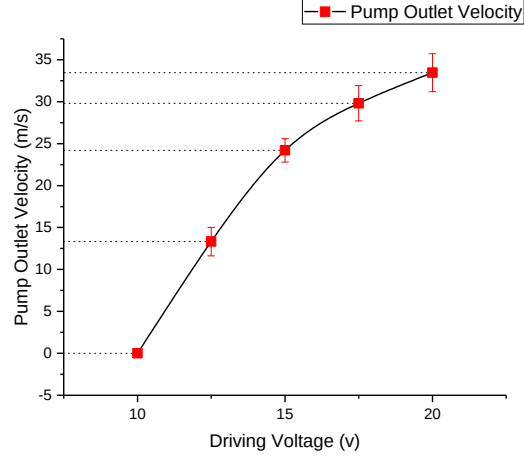


Figure 5.7: Pump Outlet velocity with respect to driving voltage at resonant frequencies. Velocities were measured five times for each driving voltage and the results are shown as error box.

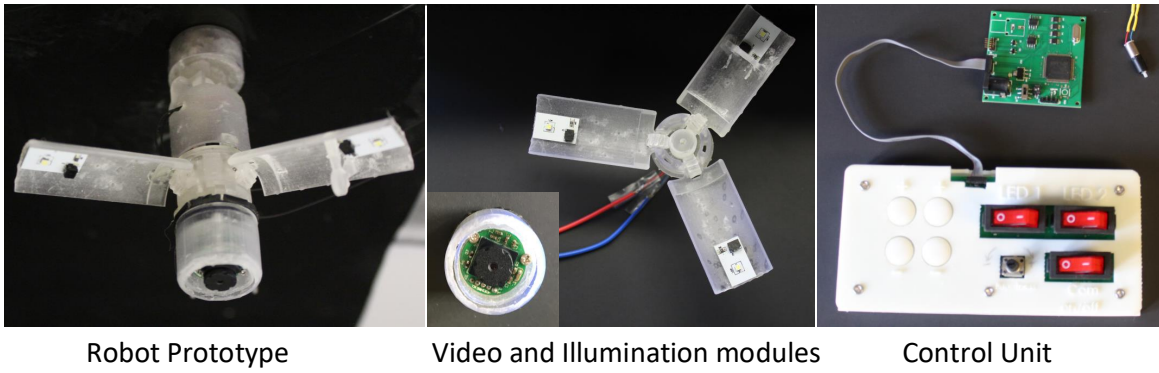


Figure 5.8: (a) Manufactured camera robot prototype; (b) Video module and Illumination module; (c) Control Unit

cavity with average thickness of $25mm$ to $45mm$. The camera robot can be manipulated for X_I and Y_I motion as well as pan rotation around Z_I .

5.6.3 Pan, Tilt and Spread Mechanism

Pan mechanism is actuated by magnetic coupling of EPM in holder and IPM in robot. Diametrical magnetization of magnets enables a smooth controllable 360 degrees pan motion. Tilt mechanism is designed to provide proper view angle for camera and illumination modules to improve the video quality. This mechanism is able to tilt to 45 degrees in maximum and

stall the camera in proper angle. Considering the 60 degrees lens's field of view, the camera robot is able to cover -75 to $+75$ degrees inside the abdominal cavity as shown in Fig.5.9.

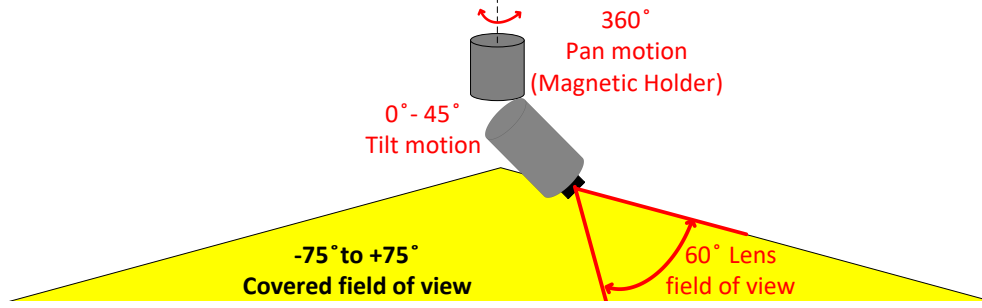


Figure 5.9: covered view angle of camera robot inside the abdominal cavity.

The wing spread mechanism is able to spread and hold the LED wings up to maximum 70 degrees to provide proper illumination. Also, it spreads and retracts repeatedly and fast to actuate the lens cleaning module.

5.6.4 Debris Prevention Test

The prevention module provides a gas barrier in front of the camera lens to repel the gas streams and debris coming toward the lens during surgery. A Low profile, high speed and high pressure air pump were used for this purpose. This pump weighs 1.4 g with 1.85 mm thickness and 20 mm in diameter core part. Murata piezoelectric pump uses 15 Vp – p driving voltage and a very low current square signal output of signal generator with 25 kHz frequency. This pump provides more than 1.42 kPa static pressure with more than 0.70 L/min flow rate. The operating temperature is also 0 to 70 Celsius which make it a proper choice for in-vivo use.

The pump outlet is connected to camera through a designed flow channel inside the structure. For this experiment a stream of smoke, generated by smoke machine, was blown toward the camera lens as shown in Fig.5.10(a) when the pump is off. after turning on the pump, the stream was deviated from the camera lens and eddy current were created far from the lens as shown in Fig.5.10(b).

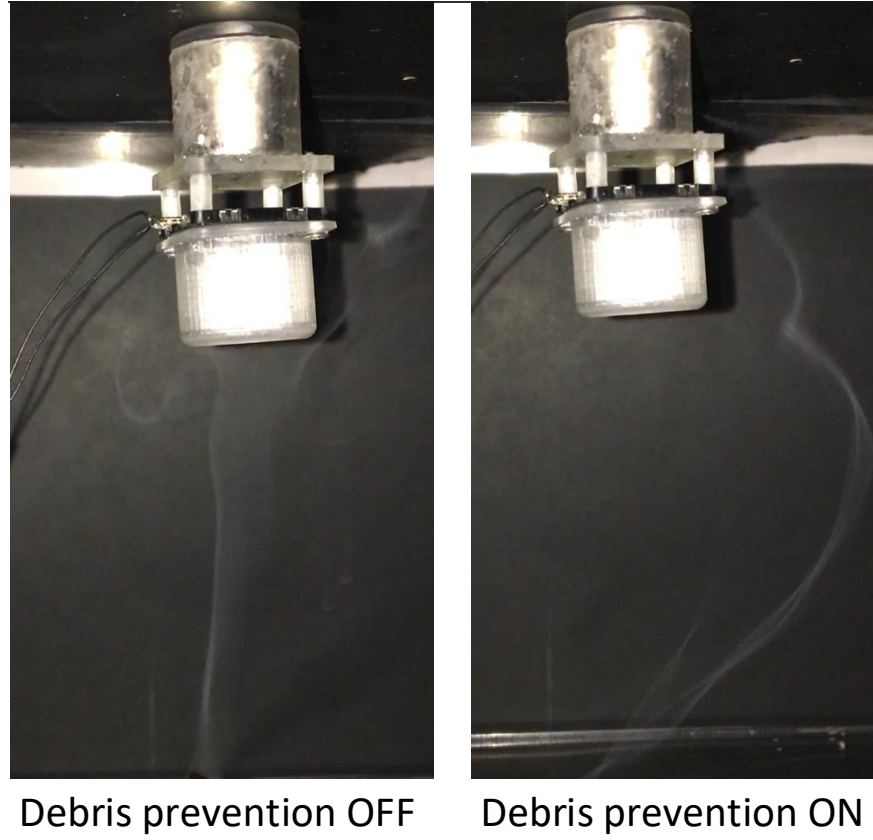


Figure 5.10: (a) Debris prevention system test on smoke stream; (b) deviated smoke stream prevents the image blurring;

Prevention module also can remove the fog created over the lens by constant flow over it. The channel were designed to blow the lens as well. The results proves that in-vivo prevention module can help deviating the streams and debris to contact the camera lens.

5.6.5 Lens Cleaning Test

Cleaning system was manufactured by 3D-printing and it holds video capturing module inside it as shown in Fig.5.10(c). Only one wing is attached to show the mechanism clearly. Two compression spring with 1.4mm outer diameter and 0.2mm wire size were used to return the wiper to its initial position after each wing retraction. Springs are mounted over 0.7mm circular tracks to make the procedure smooth.

The camera lens is covered with a clear thin cover treated with nano-tech hydrophobic solution to reduce the surface tension and ease the cleaning procedure. Standard gauze was

used as wiper tissue to clean the lens ty using the wing spread mechanism. SynDaver kidney model is used to capture the video. By using humidifier, the lens was covered with fog to replicate the abdominal cavity fogging situation. Fig.5.11 shows the images before and after the wiping mechanism for fogged lens. Another experiment were conducted for water droplets over the lens. This system is able to clean the lens off the fluid droplets as well. The results of experiment clearly shows that the cleaning module is able to remove contamination over the camera lens by wiper.

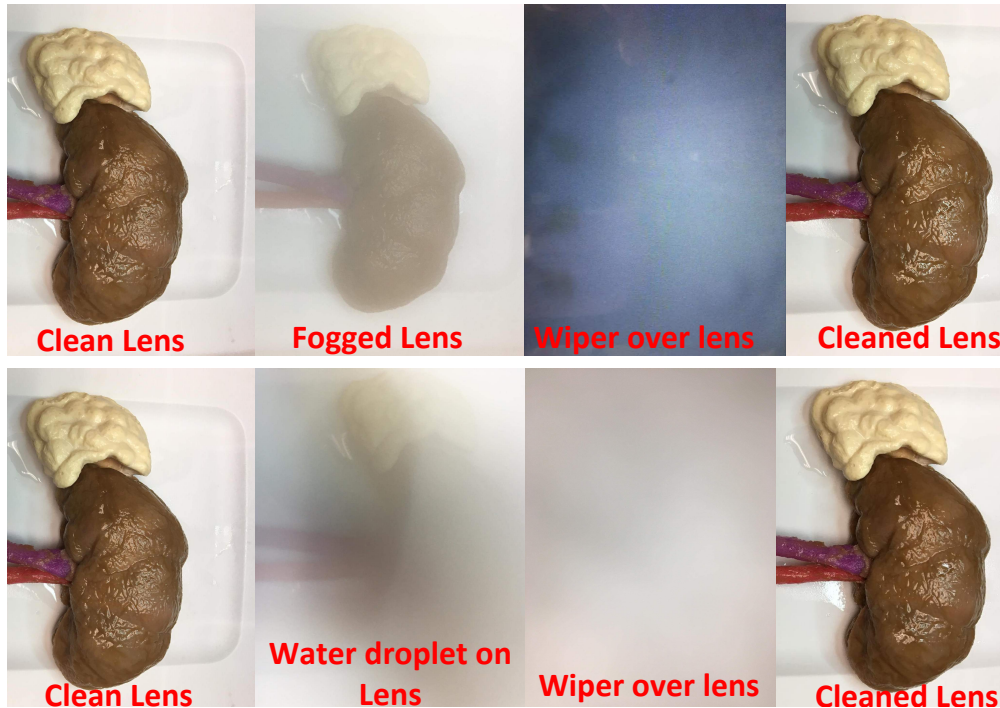


Figure 5.11: Cleaning module test for impaired image with vapor and water.

5.7 Summary

In this chapter a camera robot system with the aim of in-vivo cleaning and debris prevention has been designed and manufactured. This robot can provide debris-safe video feedback for the surgeons which will result in decreasing the surgery time and cost, will enhance the surgery work flow and reduces anxiety of surgical team. The camera robot is able to adjust the illumination of surgery area by changing the wing spread angles and can provide higher quality illumination compared to fixed, near camera LEDs.

In our future work, a smaller version of this camera system will be provided which can reduce the size and weight of the robot. Illumination system will be improved to provide shadow-less surgical area which can be an improvement for surgical videos. The camera robot system will be used for laparoscopy surgery on alive pigs.

Chapter 6

Conclusions and Future Work

This research aims at developing a magnetic actuated fully insertable robotic camera system to enhance procedures of single incision laparoscopic surgery. The design objective is to feature unified fixation, translation, and rotation functions for the camera actuation mechanism by controlling an externally generated magnetic field from a stator. The stator includes permanent magnets and coils, utilizing a closed-loop control system for achieving automatic orientation control of the camera. The actuation capabilities of the proposed design were thoroughly evaluated by experiments.

6.1 Conclusions

To design a practical actuation mechanism based on the second prototype, a hybrid stator that consists of three permanent magnets and two coils was developed for reliable camera motion control with a simple control algorithm. Experiment investigations indicated that the hybrid stator with the line-arranged rotor design can provide reliable anchoring, translation, 360° pan motion control, and up to 80° tilt motion control. The pan and tilt motions can be simultaneously controlled in a decoupled way. Therefore, this successful prototype was considered as the final design.

Grounded on the final design of the actuation mechanism, a closed-loop control system was developed for automatic control of the camera. The main components of the control system include the pan motion mechanism in the stator, the central EPM adjusting

mechanism, the abdominal wall thickness sensing system, and pan/tilt motion sensing system. The experimental results indicate that The camera orientation control can achieve 360° continuous pan motion with 0.49° control accuracy, and at least 75° tilt motion with 0.67° control accuracy. The combined orientation control in three-dimensional space demonstrated less than 1° control accuracy.

An initial prototype of a transformable *in-vivo* robotic camera system integrated with an optimized illumination system for single-port laparoscopic surgery. This prototype includes an EACU, an *in-vivo* robotic camera, and a user controller. The key innovation of this work is to introduce freeform optical lenses into the *in-vivo* robotic camera for controlling light beams from the LEDs to achieve optimized illumination. To design freeform lenses, we propose an effective ray-mapping based numerical computation method for the freeform optical lens design. The resulted illumination system features greater than 97% illuminance uniformity, greater than 80% optical efficiency, and greater than 14.323 lx illuminance on the target area when the camera-to-target distance ranges from 50 mm to 100 mm. In the experiment, we demonstrate controllability of the robotic camera, performance of the optimized illumination system and effectiveness to apply the robotic camera in a suturing task by using a simulated abdominal cavity.

6.2 Future Work

This dissertation presents a promising magnetic actuated fully insertable surgical camera to help surgeons for improving procedures in single incision laparoscopic surgery. The innovative design features a unified and automatic actuation mechanism which can anchor, navigate, and rotate a laparoscopic camera wirelessly with a capsulated body. The experiment investigations demonstrate reliable, practical, and accurate control of the camera robot. To move a further step towards a real surgical instrument that can be commercialized, the future work includes:

Development of the camera delivery system. To insert the camera into an abdominal cavity and position the camera under the magnetic coupling of the stator, a miniature flexible continuum manipulator that can be fitted in a standard trocar will be developed.

Development of autonomous surgical device tracking algorithm. During a surgical procedure, a laparoscopic camera has to be continuously adjusted by surgeons in order to achieve high quality visual feedback. In the future work, besides controlling the camera system by a user interface, a surgical device tracking algorithm will be developed that enables our robotic camera system to autonomously adjust the camera orientation and center the surgical instruments in the imaging feedback.

Ex vivo and *in vivo* tests. In the current stage, the camera system has been tested by simulated abdominal wall environments. To further investigate the actuation capabilities in real application scenarios, *ex vivo* and *in vivo* tests will be conducted by using porcine abdominal cavities.

Another step in improving the current design is to augment the laparoscopic vision with the estimated pose information of the camera. The depth information and 3D surfaces in the surgical environment could be restored from the laparoscopic vision aided by the camera motion information using the structure from motion technique. Moreover, the reconstructed 3D surgical environment could be registered into the world coordinate system through the camera pose estimation results, which facilitates intra-operative surgical planning and navigation.

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Vita

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