Fall 12-1-2010

IMPROVED MOBILE WIRELESS IN VIVO SURGICAL ROBOTS: MODULAR DESIGN, EXPERIMENTAL RESULTS, AND ANALYSIS

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IMPROVED MOBILE WIRELESS IN VIVO SURGICAL ROBOTS:
MODULAR DESIGN, EXPERIMENTAL RESULTS, AND ANALYSIS

By

Jeff A. Hawks

A DISSERTATION

Presented to the Faculty of
The Graduate College at the University of Nebraska
In Partial Fulfillment of Requirements
For the Degree of Doctor of Philosophy

Major: Interdepartmental Area of Engineering

Under the Supervision of Professor Shane M. Farritor

Lincoln, Nebraska

December 2010
Laparoscopic surgery results in superior patient outcomes as measured by less painful recovery and an earlier return to functional health compared to conventional open surgery. However, the difficulty of manipulating laparoscopic tools from outside the patient’s body generally limits these benefits to patients undergoing relatively simple procedures. The use of miniature \textit{in vivo} robots that fit entirely inside the peritoneal cavity represents a novel approach to laparoscopic surgery. These robots enable more complex laparoscopic procedures, increasing the number of patients that benefit from laparoscopic surgery.

This thesis describes recent work focused on developing a modular wireless mobile platform that can be used for surgical vision and task assistance. The modular platform can contain a variety of tools. Design details, experimental results, and analysis of new robot prototypes (cautery, clamping, staple, sensory feedback, etc.) are presented. A biopsy tool is also redesigned from previous work. Finite element analysis and experimental results are used to analyze the grasper design, which successfully removed a liver tissue sample. Tools can be removed and exchanged in a few minutes allowing a surgeon to equip the robotic platform with the appropriate tool for the desired surgical assistance. Also, a retractable cautery device is developed. Lab experiments successfully
cut and cauterized objects simulating blood vessels while another mobile platform cooperatively held the sample for cutting. Finally, visual and physiological sensory feedback packages are used to provide surgeons real time data from within the abdominal cavity.

These types of self-contained surgical devices are much more transportable and much lower in cost than current robotic surgical assistants. Furthermore, such devices can be carried and deployed by non-medical personnel at the site of an injury. Moreover, a remotely located surgeon could then use these robots to provide critical first response medical intervention irrespective of the location of the patient.
Acknowledgments

Completing this work would not have been possible without the support of my fellow collaborators and mentors. At UNL, my advisor, Dr. Shane Farritor, played a significant role during the final push to complete this work. I thank him for his patience and understanding in navigating the obstacles that surfaced during the final few months. I also want to thank him for an opportunity to begin teaching, and for the advice he’s shared about future opportunities.

Dr. Stephen Platt, who is currently at the University of Illinois-Urbana/Campaign, was an outstanding supporter and mentor. I thank him for providing me an opportunity to get started on this research during his tenure at UNL. Also, I thank him for his patience and understanding during the many emails, phone calls, and visits after his move to Illinois. His feedback regarding journal manuscripts was extremely helpful. Likewise, his advice will continue to help me in my future endeavors.

My finite element analysis would not have been possible without the direction and support of Dr. Linxing Gu. Her understanding of the Abaqus software allowed me to perform the analysis necessary to help complete this work. I also want to thank Dr. Farritor, Dr. Platt and Dr. Gu for taking the time out of their very busy schedules to read this dissertation.

None of the surgical analysis and in vivo testing results would be possible without the help of Dr. Dmitry Oleynikov at UNMC. His guidance of the surgical perspective of the design was crucial to the development of this research. Also, valuable surgical data was gathered during a collaborative surgery with the help of Dr. Mark Rentschler and surgeon Dr. Jonathan Schoen at the University of Colorado. Mark provided a great starting point for this research, and I thank him for his advice and experience as I continued on with his work. Also, Jacob Kunowski at UIUC was a great help in providing circuit boards and parts during the final year.

I want to thank my mother and father for giving me the opportunity to pursue my endeavors with their steadfast support. I credit my father for initiating my love of design and problem solving. And finally, I would like to thank Ashley for her loving support and understanding of the time commitments and late nights as I pursue my career and we begin a family together.

JEFF A. HAWKS
Lincoln, Nebraska
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Introduction

Surgical practices have changed radically over the past twenty years. Conventional practices, which require large incisions to gain direct access to the surgical site, have given way to minimally invasive surgery (MIS). MIS has become popular for relatively simple surgical procedures. The large incisions are replaced with small incisions that allow long, slender tools to gain access to the surgical site. The use of robotic technology has recently become a significant factor in improving this type of surgery. Robots have been developed to assist surgeons in performing these procedures.

Specifically, the robots discussed in this thesis were designed to provide surgical vision and task assistance from within the abdominal cavity during laparoscopic surgery. The mobile robotic platforms developed during this research are controlled wirelessly. The modular design of the robotic platform facilitates the rapid development of a variety of configurations of these robot prototypes. For example, payload variations that allow the mobile robot platform to collect tissue samples, communicate physiological sensor
readings, manipulate organs, provide visual feedback, dissect tissue, and stop bleeding are all discussed throughout this thesis.

Extensive analysis of the biopsy grasper design parameters is provided for the design of a biopsy grasper for collecting hepatic liver tissue. In the United States alone, more than 20,000 people die each year as a result of complications arising from chronic liver diseases and the associated fibrosis and cirrhosis [1]. Furthermore, the liver is the most frequently injured abdominal organ in motor vehicle accidents and falls from heights.

Finally, in vivo tests were performed using a porcine animal model because the size and position of the abdominal organs in a porcine animal model are similar to that of a human. The results of these in vivo surgical tests are presented in this thesis. The use of miniature, wireless, mobile robotic platforms can provide surgical task assistance through a single incision, potentially reducing patient trauma and recovery time compared to current surgical practices.

These miniature, wireless in vivo robots also have the potential to be field-configurable for specific incidents and applications. These types of low-cost and easily transportable robotic devices could eventually become standard equipment carried and deployed by non-medical personnel at the site of an accident or injury, providing support for remote first-response medical care. Furthermore, a remotely located medical team could use these devices to deliver a rapid therapeutic response and continually monitor physiological parameters prior to and during transport without cumbersome external connections.
Chapter 1

1 Surgical Robots Background

1.1 Minimally Invasive Surgery (MIS)

Conventional surgical practice requires the on-site physical presence of highly trained medical teams. Large incisions are typically made so the surgeon can see and place instruments directly into the operating site. These invasive techniques lead to more patient trauma and longer recovery times compared to minimally invasive surgery (MIS). It is widely accepted that the trauma inflicted while gaining access to the operating site often causes additional injury to the patient, resulting in more pain, longer recovery times, and increased morbidity [2-4].

MIS reduces this collateral trauma. Laparoscopy is MIS performed in the abdominal cavity through small tool ports, called trocars, with long slender tools. Carbon dioxide gas is pumped into the abdominal cavity to create a larger volume of space for surgical
operation and visualization. A surgeon operates the tools while viewing the surgical field using a laparoscope, which is also inserted through a trocar.

Surgical procedures performed through a few small trocars placed in the abdominal wall greatly reduce patient recovery time and trauma when compared to conventional surgical practices. Studies clearly show that laparoscopic procedures result in shorter hospital stays, less pain, faster return to the normal activities of daily living, and improved immunologic response. In particular, minimally invasive surgery reduces postoperative hospital stays to just over a day compared to over a week for standard procedures [5]. However, laparoscopic procedures are difficult to perform. Operating with long, slender tools that are placed through trocars is not an intuitive process, and requires years of training before they are mastered by surgeons.

Even the most gifted surgeons are subject to human limitations [6]. For example, the effects of fatigue and hand tremor limit the precision of motion. To further complicate matters, a surgeon’s skills are limited by poor access to the surgical site. To illustrate, laparoscopic surgery is often termed “chopstick surgery” because the operations must be performed using long, thin instruments. These instruments are inserted through trocars, which significantly reduce the dexterity of the instrument tip. Haptic and tactile feedback is also eliminated during minimally invasive surgery [7]. Friction between the laparoscopic instrument and the seal of the MIS port as well as the forces exerted on the port by the abdominal wall greatly reduce the surgeons’ ability to feel the tissue [5]. Therefore, contact force must be estimated by visual observation of tissue deformation and color change. Furthermore, visualization is typically impaired due to limited depth perception, field of view, and difficulty coordinating camera motion. Consequently, the
overwhelming success of laparoscopy in relatively simple procedures (e.g., gallbladder removal) has not been replicated in more complex procedures [2].

1.2 Robot-assisted MIS

There are many important factors driving the acceptance of surgical robotics. Perhaps the most obvious factor is the ability of robotic systems to significantly improve a surgeons’ technical capability [8]. This can be done by making existing procedures more accurate, faster, or less invasive. Robotic assistance also makes it possible to perform otherwise infeasible operations when human limitations make intricate procedures too risky. Moreover, the goal of robot-assisted MIS is not to replace surgeons, but to provide surgeons with a new set of robust tools that extend their ability to treat patients.

The development of robotic surgical tools has helped to remove some of the limitations and reduce complications associated with traditional manual laparoscopy. The first robots designed to assist during minimally invasive abdominal surgery (e.g., LARS and AESOP) appeared in the mid 1990s [9,10]. The most commonly used robot today is the da Vinci surgical system, which received FDA clearance for sale in July of 2000 and is made by Intuitive Surgical. It is currently the only system commercially available, although other systems such as AESOP are still used. The da Vinci is a tele-robotic system that is controlled by a surgeon at a console. It operates in a master-slave relationship with the surgeon, where robotic arms hold the camera and instruments. Advantages of such robots include reduction of tremor, additional articulations in surgical instruments, corrections for motion reversal, and motion scaling. However, all of these robots are situated outside the patient and thus remain subject to the dexterity limitations
imposed by the use of long tools inserted through small incisions. Some studies suggest that current externally-situated robotic systems offer little or no improvement over standard laparoscopic instruments in the performance of basic skills [11,12].

The da Vinci surgical system is also cumbersome, the tool changes are difficult [13,14], and significant set-up time and operational space are required [15]. Size is an important design characteristic for surgical robots. Operating rooms are usually small, and large robots can take up too much space. The movement of the large external arms also makes direct access to the patient difficult, and their motion must be limited to avoid internal and external collisions [16]. A limited range of motion for the robotic camera can still result in obstructed or incomplete visual feedback. Likewise, the robots’ high cost and extensive support requirements make them unavailable to many hospitals [17].

Current efforts in robotic surgery are focused on developing next generation robots that improve mobility and sensing capability while reducing complexity and cost compared to systems like da Vinci. For example, researchers at the University of Hawaii-Manoa are working on a compact portable teleoperated robot for laparoscopic surgery [18]. Likewise, the Medical Robotics Group at the University of California-Berkeley has built and tested a prototype laparoscopic robot with force and tactile feedback for telesurgical applications [19,20]. Similarly, the Carnegie Mellon University Robotics Institute is developing intelligent microsurgical instruments to electronically cancel tremor in handheld surgical tools [21,22]. Prototypes of new endoscopic tools with force and tactile feedback are being created at the Bio-Robotics Laboratory at the University of Washington [23].
Other work is focused on developing robots in which all or most of the device enters the patient’s body. The simplest such mechanisms have been maneuverable endoscopes for colonoscopy and laparoscopy [24,25]. These devices possess actuators to rotate the endoscope tip after it enters the body. Other \textit{in vivo} robots have been developed to explore hollow cavities (e.g., the colon or esophagus) with locomotion systems based on ‘inch-worm’ motion that use a series of grippers and extensors [26], rolling tracks [27], or rolling stents [28]. These devices apply radial pressure to the walls of the hollow cavities they explore, and thus cannot operate in the open space of an insufflated abdomen.

Another approach relies on an untethered pill that is swallowed and passively passed through the gastrointestinal (GI) tract. One such commercially available device [29,30] returns thousands of images as it naturally moves through the GI tract. However, because the device navigates passively, it cannot be directed to image a particular location, and the precise locations of the images returned are unknown. Combined with the large number of images, the use of this device for specific diagnosis is difficult. Other similar devices are now available [31,32]. Dario et al. have recently described an endoscopic pill with an active locomotion system that uses legs to push against the gastrointestinal walls [33,34] and a clamping system that uses shape memory alloys [35]. Currently, this device is still in a conceptual development stage.

A proof-of-concept design of an \textit{in vivo} stereoscopic imaging system has been described by Miller et al. [36]. A second generation single camera pan and tilt prototype based on this initial concept is described in Hu et al. [37], and is currently being evaluated in \textit{ex vivo} and \textit{in vivo} tests. Finally, the HeartLander robot employs a suction-based drive to move across the surface of the beating heart [38,39]. Prototypes have
demonstrated successful prehension, turning, locomotion, and dye injection in a porcine model.

One of the main goals of minimally invasive robotic surgery is to improve the surgeon’s sensory feedback and tool manipulation capabilities. The previously mentioned examples offer intuitive ways to improve vision of the surgical field, eliminate hand tremor, or perform dexterous operations inside small cavities. In addition, the use of surgical robots, such as da Vinci, also enables a surgeon to control more than two arms during complex procedures, which can improve the efficiency of the procedure. Put simply, surgical robotic devices extend and enhance the capabilities of a surgeon’s hands.

Robotic surgery is a rapidly growing area in the global health care industry [40]. Developers argue that robotic technology saves everyone money in the long run. To illustrate, robot-assisted minimally invasive surgeries allow patients to go home and return to work earlier. There are fewer complications compared to traditional surgery, which reduces cost for hospitals and insurers. Clinical evidence is beginning to support the developers’ claims. Most robot-assisted MIS patients are back at work after two to four weeks, compared with six to eight weeks for open-chest patients. Also, the average cost of surgery declined by 5% to 10% compared with conventional open-chest surgery [40]. Furthermore, by miniaturizing and reducing the costs of surgical robot systems such as da Vinci, surgery costs will continue to decline.

1.3 In vivo Laparoscopic Robots

Most of the in vivo robots described previously require narrow cavities or natural processes for their mobility systems to function, and/or external connections for actuation, power, and tool control. They are designed to function in locations other than
the abdominal cavity. The open environment of an insufflated abdomen during laparoscopic surgery is incompatible with many of these approaches. By placing robotic tools inside the abdominal cavity near the surgical site, critical issues related to ergonomic limitations, reduced dexterity, and limited perception are mitigated. While *in vivo* laparoscopic robots currently lack some of the precise control provided by systems such as da Vinci [17], they have been useful in providing vision and task assistance.

Several research studies analyze the advantages of placing robots for surgical assistance inside the abdominal cavity. For example, test results show a significant decrease in the time required to complete basic laparoscopic tasks when using the *in vivo* camera robots. These results suggest that video feedback from miniature *in vivo* robots is at least as good as, and perhaps superior to, that from the laparoscope for these tasks [17].

### 1.4 Wireless *In vivo* Surgical Robots

While research efforts discussed in previous sections of this thesis described technology that reduces the size of surgical robot systems, tethered support hardware is needed to power, control and monitor feedback from many of these devices. For example, the HeartLander [44] was connected to a suite of support instruments through a flexible tether. Offloading functional components of the robot, such as motors and power supply, was a convenient way to miniaturize the robotic device. For the HeartLander, maneuverability was not significantly hindered by the tether because the operating space of the robot is a single continuous volume. Tethers present in an insufflated abdominal cavity have a tendency to become tangled or cumbersome.

Many endoscopic robots must use a similar tether to transmit power and control signals. When navigating the bowels, the tether causes limitation of working area and
mobility. If the tether is constructed of a rigid material, it may also injure the intestine. Therefore, researchers have developed micro-robots based on wireless communication [45], power transmission, and video transmission [46].

Another wireless device, the NORIKA 3 [7], was propelled in the gastrointestinal tract by exploiting the force generated by external electromagnetic fields, which can be tuned by a joystick. The capsule incorporated a charge-coupled device (CCD) camera and some drug-delivery modules for localized therapy. This system did not incorporate on-board intelligence and was essentially a wireless teleoperated device rather than a reactive and adaptive system.

1.5 Mobile Wireless In vivo Surgical Robots

The use of miniature in vivo robots that can be inserted through a small incision and fit entirely within the peritoneal cavity represents a novel approach to laparoscopic surgery. Our previous work has focused on developing a family of in vivo fixed-base and mobile robots, and demonstrating that they can successfully operate within the abdominal cavity. These robots have been used to enhance the ability of laparoscopic surgeons to visualize the surgical field [41,42], and to obtain tissue samples during a single-port liver biopsy in a porcine model [43]. However, these robots relied on tethers for power, control, and data transmission, and each individual robot was designed for a specific task. Robots presented in this report are wireless, and their modular design reduces development costs. Likewise, the in vivo robots presented in this thesis are small, relatively inexpensive, and easily transportable, hopefully making this technology to be more widely adopted.
Chapter 2

2 Robotic Platform Design

2.1 Previous In vivo Mobile Robots

During previous research efforts, the robot body was specifically designed to accommodate the functions of the robot. For example, smaller robots were designed when only a camera was housed inside the robot body. Meanwhile, a slightly larger robot was designed in order to allow more space for housing both a biopsy grasper and video camera. While the robots appear to be similar, their construction and design were indeed unique and specific towards the surgical assistance they were designed to provide.

The robots were machined out of aluminum. Due to machine tolerances, motors and other components were typically glued to the robot body. Consequently, the robot components were very limited in terms of adjustments or reconfiguration. As prototypes progressed, new robots were built from scratch. The time and cost required for this development led to only a few robots built and tested each year.
The general design of the wireless mobile robotic platform builds upon our earlier work developing tethered *in vivo* mobile robots [47,48], shown in Figure 2-1. Previous designs consisted of a cylindrical inner housing, two wheels that slipped over the housing, and a tail that could be collapsed into the wheel treads when the robot was inserted or retracted through a trocar. The wheels allowed forward, reverse, and turning motions, and the tail prevented counter-rotation of the robot body when the wheels were turning. Depending on the desired function of the robot, the tail could be as simple as a stainless steel coiled spring. However, in the case of a biopsy grasper arm or other tool that protruded perpendicularly from the robot body, the tool itself could be designed to satisfy the functions of a tail.

Improvements to previous designs include less expensive manufacturing techniques. A modular design approach reduces development costs, and allows robots with different task assistance goals to be produced without completely redesigning or rebuilding the major components of the robot platform. Furthermore, these improvements offer significant improvements in the functionality of mobile wireless *in vivo* surgical robot. We also anticipate these robots to be single-use disposable devices to mitigate challenges of re-sterilization, component fatigue, and battery recharging.

![Figure 2-1: Tethered In Vivo Mobile Robot (Rentschler, 2006)](image)
2.2 Modular Base Design

A wireless modular robotic platform has been developed to advance the ability to use \textit{in vivo} robots for diagnostic and surgical support. The body of this robot is approximately 20 mm in diameter and 100 mm long. Two wheels, each driven independently by a permanent magnet direct current motor, allow forward, reverse and turning motions.

The construction of the robot inner housing differs in several significant ways compared to previous efforts. The housing itself is modular, consisting of two halves with each half comprised of two clamshell-like pieces with a semi-circle profile. Stereolithography prototyping (rapid prototyping) techniques are used to manufacture the housing components out of ultraviolet-cured (UV-cured) PolyJet FC720 Clear 83D \cite{49}. The material withstands 15-25\% elongation before fracture and has a tensile strength of 60 MPa. The heat deflection temperature of 44° Celsius is above body temperature, and the tolerances on the PolyJet can be less than 0.1 mm.

Early concerns surrounded how the material would stand up to the wet and warm environment of the abdominal cavity. Another concern was the ability to maintain a rigid housing with tolerances that would allow a fully functional robot. However, lab experiments tested the material’s rigidity when placed in moist environments.

The advantages to these manufacturing techniques are extremely significant. Parts are manufactured in a few days instead of weeks, and are produced at nearly a quarter of the cost of aluminum parts used in previous work. The material is much lighter than aluminum, and consequently, the weight of the robot is greatly reduced, which reduces stress on abdominal organs and tissues. Another important aspect is that the PolyJet
material is able to withstand surface finishing such as wet sanding. This allows smooth, low friction areas to be manufactured without adding fittings or bearings.

Figure 2-2 shows a basic schematic of modular robot base. Each half of the inner housing has a specific purpose. The modular half of the body houses the power plant (i.e., battery), the master control microprocessor and radio frequency (RF) communication electronics, and a permanent magnet direct current wheel motor. The second half of the body also houses a wheel motor, but the majority of the volume is a dedicated payload space (e.g., biopsy, sensors, etc.). This design approach provides for a common mobile platform that could be easily re-configured for a variety of task-specific applications.

The clamshell pieces of the modular side of the robot are shown in Figure 2-3. The benefits of the PolyJet manufacturing process are clear. For example, the complex geometries shown below are extremely difficult and expensive to machine out of metal. Cavities for holding the battery and motor can be built to fit specific components. The motor assembly fits into a channel, and the flexibility of the plastic allows it to snap into place, eliminating the need to glue the motor to the robot body.
The clamshell design allows components to be placed into the two halves as shown and then placed together. Consequently, assembly is much easier and faster compared to earlier robot designs. Holes for assembly screws are tapped on one side of the clamshell pieces so that machine screws will fasten the two halves together. Moreover, components can easily be replaced without damaging the robot housing. The modular design allows the robot to be assembled, disassembled, and reassembled multiple times without manufacturing new pieces, making these robots ideal for the changes and adjustments during research and development.

The control board and battery fit into a tray designed to keep it from sliding and twisting, which helps eliminate the possibility of wires disconnecting. The diameter of the housing is 16 mm, and is driven primarily by the size of the battery. The wall thickness of the robot body is approximately 0.2 mm in the location of the battery. Before the robot is assembled, the thin wall is fragile and can easily be fractured. However, once the battery is placed inside the robot, the battery reinforces the wall and the robot housing becomes very durable. Wall thicknesses in other areas of the robot are at least 0.75 mm, and therefore, do not need any form of reinforcement.

Another example of the complex geometries possible with the PolyJet construction is the wire conduit that is specifically designed for wires to pass over the top of the battery.
Figure 2-4 shows the modular half of the robot during assembly. These wires connect the battery to the control board. The conduit space also allows room for wires that connect the control board with any motors or circuit boards that may be located in the payload side of the robot. Again, the conduit is designed for the wires to fit tightly when the clamshells are fastened together so that wires would not twist and disconnect during assembly or operation.

![Figure 2-4: Modular Half of Robot during Assembly](image)

### 2.3 Wheel Design

Wheeled mobility is an important feature of the modular robotic platform. To illustrate, the robots can be positioned to provide visualization or surgical assistance nearly anywhere within the abdominal cavity. However, internal abdominal organs and surfaces are highly deformable and very slick, and the constitutive relations describing wheel-organ interactions are quite different compared to those of terrestrial terrains. We have previously explored the nature of wheel-tissue interactions through analytical modeling and empirical analysis of experimental results [47]. This work led to a general helical tread wheel design that has been shown to provide mobility across abdominal organs and surfaces without causing tissue damage [43]. Additional work using finite element analysis [48] has provided a better understanding of how changes in robot mass and wheel geometry affect robot mobility.
Based on these results, wheels, shown in Figure 2-5, were designed for the current application. These wheels are 20 mm in diameter with 9 helical grousers arranged in a corkscrew pattern. The grousers have a depth of 1.5 mm, a thickness of 1.75 mm, and are spaced at 7 mm intervals. The pitch angle is 10.6° so that a minimum of two grousers are in surface contact at all times to help ensure a smooth motion profile. The wheels are bored to accommodate the robot body housing, and have an inner diameter of 16.25 mm with a wall thickness of 0.375 mm. The grouser treads provides additional mechanical support to the thin walls. At the end of each wheel is a small ball-like feature that facilitates handling of the robot by the surgical team using laparoscopic tools during insertion and retraction.

Figure 2-5: Schematic of Wheel

The wheels are manufactured using the same stereolithography techniques and material used for the robot body housing. The PolyJet material used is wet sanded in order to achieve a smooth surface on the outside of the modular robot body and the inside of the wheels. The wet sanding eliminates the need to have any type of bearing between the robot housing and the wheel. The surface finish has low enough friction properties to serve as a bearing for the wheel during operation.
Each wheel is actuated by a 6 mm diameter permanent magnet direct current (PMDC) motor with a 256:1 gear ratio manufactured by MicroMo. The gear ratio allows a high torque output and steady speeds. Using a smaller gear reduction, such as 64:1, would also be acceptable for the wheels, and it would increase the speed of the robot. A spur gear on the motor shaft couples the motor to a spur gear and bearing assembly mounted on the end of each wheel.

2.4 Electronics and Wireless Communication

The robot master control circuit board is shown in Figure 2-6. This is a custom designed double-sided surface mount printed circuit board (PCB) that incorporates an RF transceiver, a multi-channel integrated circuit (IC) that can drive both voltage-controlled and constant current-controlled actuators or other components (e.g., permanent magnet DC motors; voice coils), and a master microprocessor control unit (MCU). A Hall effect sensor switch allows the board to be turned off by placing a magnet near the circuit board. This is ideal for packaging or storing a fully assembled robot without draining the battery power prior to operation.

![Figure 2-6: Top (left) and Bottom (right) View of Master Control Circuit Board. (© IEEE 2009)](image-url)
The metal casing of the battery provides a natural attachment area for a magnet, and it is close enough to the circuit board to activate the switch. Concerns of magnetizing the PMDC motors over long storage periods were dismissed through lab experiments. During these experiments, the robot was powered on and off daily for a period of two months. A programmed initialization routine tests the motors during the power up procedure, which indicates that the robot was functioning properly. Over the course of the two months, the motors functioned properly.

The wireless communication unit is built around a Nordic nRF2401A 2.4 GHz ISM band single-chip radio transceiver. This low-power, fully-integrated transceiver is capable of error-checked data rates up to 1 Mbps in burst mode. Because there are 125 receive/transmit channels, multiple robots can be used simultaneously without interfering with one another, which fits into the vision of cooperative robots working together inside the abdominal cavity. The transceiver is configured using a 3-wire serial interface to the control board MCU. A differential to single-ended matching network based on the Nordic nRF2401 reference design is used to accommodate a single-ended connection to a 50 Ohm chip antenna (LINX ANT-2.45-CHP).

A Toshiba TB6557FLG driver IC is used to provide up to six H-bridge (two constant current-controlled and four voltage-controlled) output drivers. This IC is also configured using a 3-wire serial interface to the MCU. Two of the voltage-controlled outputs are generally dedicated to controlling the PMDC wheel motors. A third voltage-controlled output is used in the robot payload area to drive any potential tool motor. Other types of actuators and components can be accommodated with the current design. For example, it is anticipated that the constant current-controlled outputs will be used in a future robot to
control a voice coil actuator as part of an adjustable focus camera system. It is also likely that a light emitting diode (LED) lighting system or heating element, such as a cautery tip, can be controlled similarly.

The master MCU is a PIC16LF767. This low-power processor includes a 10-bit analog-to-digital converter with up to 11 input channels, three independent pulse width modulation modules, and extensive power management features that can minimize power requirements. In its current configuration, the processor operates at 4 MHz, although a variety of lower or higher frequencies can be used. The MCU is responsible for configuring the various robot peripherals (e.g., transceiver, driver IC, and sensors), reading the sensor data, controlling actuators, transferring data to and from the communication module, and various other housekeeping tasks. The MCU control program is common across all robot variations. However, various application specific routines can be turned on or off to improve performance by using a MCU input pin to set internal flags. For example, because there is no need for certain payload variations to execute code related to reading and transmitting sensor data, this portion of the code can be disabled without reprogramming the entire device.

Furthermore, the control board is designed to allow in-circuit programming, which allows the robot to be easily reprogrammed prior to assembly. Circuit reprogramming is essential in the modular design and development of these robots. For example, circuit boards can be produced in mass quantities without programming the circuit board until the specific function of a robot is identified. Likewise, if robot functions change during development process, the circuit boards can be reprogrammed and tested without constructing new boards.
All on-board power is provided by a single high power 185 mAh lithium organic cell battery (Tadiran TLM-1520MP). The battery has a length of 20 mm and a diameter of 14.8 mm, which is the driving force of size limitations of this robotic platform. The open circuit voltage is approximately 4.05 V, with a maximum discharge current of 1 ampere continuously to 2.8 V. Furthermore, this battery has sufficient energy density to operate the robot for nearly 2 hours with all motors running continuously, and a stationary sensor robot can operate for more than 5 hours.

External control systems, shown in Figure 2-7, have also been developed to send control commands to the robot and process the in vivo data telemetry stream. These systems incorporate the same microprocessors and RF transceivers as the in vivo robots. Additional components are included as human interface devices (e.g., joysticks used to control robot wheel speed and direction and RS-232 data transfer to an external storage computer), and the microprocessor software is modified to reflect these differences.
Chapter 3

3 Biopsy Grasper Payload

3.1 Background in Laparoscopic Biopsy

Traditional laparoscopic biopsy sampling tools typically consist of a grasper on the distal end of a long flexible tube, and a handle and lever system on the proximal end. A Teflon-coated wire that runs through the tube is affixed at one end to the handle lever, and at the opposite end to the grasper. Actuation of the handle causes the wire to translate relative to the tube and actuate the biopsy grasper. Laparoscopic biopsy grasper jaws do not overlap and completely sever tissue as do the jaws of many of the biopsy punches used in conventional surgery. Laparoscopic biopsies are, therefore, typically “grasp and tear” procedures that require a relatively large amount of force to tear the sample away from the organ.

Most surgical interventions require the ability to manipulate tissue. To demonstrate the feasibility of a wireless in vivo robot to provide surgical task assistance, a biopsy
grasper and actuation mechanism payload was developed. Because the miniature motors used to actuate the *in vivo* robots cannot directly generate sufficient forces to retrieve tissue samples, significant effort was applied towards developing an actuation mechanism with a large mechanical advantage that would sever the tissue. Additional effort was devoted towards designing the mechanism and biopsy grasper tool such that multiple tissue samples could be obtained. Another key aspect of the actuator mechanism design is that it can be used with a variety of end-effectors in addition to a biopsy grasper (e.g., a clamp to hold tissue and/or control hemorrhaging).

### 3.2 Early Design Approaches

A number of alternative biopsy designs were developed and tested. For example, as described in previously published work [43], a set of standard laparoscopic biopsy forceps were modified to be actuated by a PMDC motor slider mechanism. As is typical for laparoscopic biopsy forceps, the cutting edges of the forceps did not overlap. This approach, therefore, was an attempt to replicate the “grasp and tear” biopsy procedure using a miniature robotic platform. Unfortunately, the mobile platform lacked sufficient traction to routinely and reliably tear the partially severed sample from the main tissue body.

Other design approaches included using a four-bladed mechanism similar to a tree spade to extract a tissue sample, and a garage door-type mechanism that would close over and slice a sample from a protruding edge of tissue. Both approaches are limited by frictional forces that impede the motion of the blades. Another concept that was explored was using a flat sliding blade to cut a sample from tissue that had been sucked into a
reservoir using a vacuum. Drawbacks to this approach include generating sufficient suction forces and proper placement of the robot for sampling.

After experimenting with these different design concepts, some obvious design considerations arose. First, taking biopsy samples from the edge of the liver is the most ideal location for the mobile platform. Therefore, the jaws must be designed to clamp on the edge of the liver. Second, the jaws must overlap in order to cut and tear the tissue with a minimal applied force.

The preliminary design of the biopsy grasper is shown in Figure 3-1. A super-elastic nickel titanium alloy (Nitinol) ribbon is used for the top jaw. Meanwhile, a stainless steel hypodermic tube is used for the bottom jaw. Stainless steel is a common material in surgical tools due to low cost, easy sterilization, and corrosion resistance. Although systemic contact with nickel can be toxic, Nitinol is self-passivated by a biocompatible titanium dioxide layer that has high resistance to corrosion and minimizes leaching of nickel [50].

![Figure 3-1: Preliminary Design of Biopsy Grasper Open (left) and Closed (right)](© ASME 2008)

The top and bottom jaws were machined, profiled, and sharpened so that two claw-like teeth overlap and interlock. A motor and lead screw linkage is used to actuate the biopsy grasper mechanism. The jaws close as a collar slides over the Nitinol profile and squeezed it together with the stainless steel tube. The super-elastic properties of Nitinol greatly reduce the effects of plastic deformation from cold work.
Currently, Nitinol is becoming a common material in the development of surgical robotics. For example, the HeartLander robot used super-elastic Nitinol wires to drive the crawling mechanism of the robot [44]. Another example was the use of Nitinol in a bending mechanism for a forceps manipulator [51].

### 3.3 Ex Vivo Bench-Top Tests

*Ex vivo* tests were conducted to characterize the ability of the grasper to obtain tissue samples. The biopsy grasper and actuation mechanism were placed inside the payload area of the modular robot platform. Again, a motor and lead screw linkage was used to actuate the grasper. However, during this experiment the grasper was positioned such that it would grasp onto fresh bovine liver, used as a proxy for porcine hepatic tissue, as it closed (Figure 3-2 left).

![Figure 3-2: Ex vivo Bovine Liver Tissue Sample Collection (© ASME 2008)](image)

This sampling test was repeated multiple times. As illustrated in Figure 3-2 (right), multiple samples can be collected in the reservoir formed by the stationary bottom jaw of the biopsy grasper. In each test, the overlapping jaws completely or nearly completely severed the sample from the remaining tissue mass. In the cases of incomplete severing, the degree to which the samples were cut from the main body of tissue greatly exceeded that which is typically achieved with standard laparoscopic biopsy forceps. Previous
analysis [48] indicates that the mobile robot platform has sufficient traction to provide the very small additional force that is needed to pull a partially severed sample away from the tissue.

Unfortunately, this early design lacks the repeatability and reliability necessary for *in vivo* surgical task assistance. While the thicker Nitinol cut the bovine liver tissue, it is extremely unclear whether or not it would cut living liver tissue, which is encapsulated by a stiff membrane. The use of the thick Nitinol ribbon also causes the actuation to become unreliable due to the high forces needed to close the grasper. Initial results suggest that a biopsy grasper design using a thick Nitinol ribbon would be unreliable in the *in vivo* environment.

Nitinol is not an ideal material to cut soft tissue. For instance, the material does not maintain a consistent sharp edge. Also, if a thick ribbon (0.45 mm) of Nitinol is used, a large amount of force is needed to close the grasper. On the other hand, a thin Nitinol ribbon (0.20 mm) lacks the stiffness needed to penetrate living tissue.

### 3.4 Biopsy Grasper Design

Following a lengthy series of bench-top tests, a promising general mechanism design was identified, as illustrated in Figure 3-3. The prototype biopsy grasper resembles a jaw. However, this design does not use the Nitinol ribbon as the cutting tool. Instead, sharpened blades are inserted into the top and bottom jaws. Unlike other laparoscopic biopsy forceps in which both jaws are hinged about a pivot point, only one jaw of the robotic grasper moves during sampling. The lower half of the grasper remains stationary and provides a rigid and stable base against which the upper jaw can cut.
Figure 3-3: Top View Schematic of the Biopsy Grasper (© IEEE 2009)

The profile of the top jaw is constructed out of a Nitinol ribbon (Mernry Corporation) 0.25 mm thick and 3 mm wide. It is profiled such that the grasper was normally open. A wide variety of profiles can be achieved by heat-treating the ribbon for approximately 10 min at 500 °C, followed by quenching in water. This process, commonly called a “shape setting heat treatment,” was also used in shaping Nitinol wire for inchworm locomotion [45].

A stainless steel jig, shown in Figure 3-4, was constructed to form the Nitinol to a specific profile. After heat treating, the Nitinol ribbon is glued to a fixed nylon rod insert that fits inside the bottom jaw. The nylon rod is also glued to the housing of the robot, which anchors the grasper when the collar is actuated. Finally, the fixed bottom jaw is constructed from a hypodermic medical stainless steel tube and it forms a reservoir for storing multiple tissue samples.

Figure 3-4: Jig Used for Nitinol Heat Treatment
The blades mounted to the biopsy grasper jaws are constructed of titanium nitrate coated stainless steel approximately 1.5 mm long. The titanium nitrate coating maintains a sharp cutting edge, and will penetrate the liver tissue much better than the Nitinol discussed in Section 3.3. Consequently, using these sharpened blades allow the use of a thin Nitinol ribbon for the top jaw profile, which greatly reduces the force needed for actuating the biopsy grasper.

Small plastic inserts are fixed to the top and bottom jaws, and the blades are glued to these inserts. The round blade fixed to the bottom jaw has a diameter of 3.40 mm. The top blade has a semi-circular profile with a diameter of 4.22 mm and overlaps the bottom blade when the jaw is closed. The sample is held within the bottom blade as the trailing edges of the top blade help sever the sample from the tissue.

A tissue sample is obtained by using a PMDC motor to actuate a stainless steel collar that slides over the grasper in the direction \( A \), shown in Figure 3-5. As the top jaw closes in the direction \( B \), the upper and lower cutting surfaces overlap, severing the tissue sample. When implemented on the modular robot, the biopsy arm is perpendicular to the robot body and actuator motor. Therefore, an actuation linkage inspired by our previous work [43] is used to transform the direction of translation of a lead nut driven by the motor to an axis in line with the movement of the collar. The limit of translation of the lead nut is such that the linkage approaches a mechanical singularity at the point of grasper closure, producing a large mechanical advantage.
The biopsy grasper payload provides another example of the benefits of using the PolyJet manufacturing process to construct the robot body. The payload area of the modular robotic platform is designed to perfectly fit the specific components for the biopsy grasper. Modifications to the basic modular platform are labeled in Figure 3-6. A motor housing, similar to the one housing the wheel motor, is added to house the motor to drive biopsy actuation. Also, additional assembly screws are added to reinforce the clamshell pieces against the off-axis forces present during collar actuation. For example, friction from the collar sliding against the robot body may cause the housing to split apart in the absence of these additional screws.
One design concern is safely securing the motor wires. Motor wires must go around the actuation linkage to connect with the control board in the modular side of the robot. However, the wires cannot be located near the actuation mechanism, or they may become entangled with the actuation mechanism. With the use of PolyJet manufacturing, a small wire conduit is profiled for the wires to pass along the actuation mechanism with no threat of entanglement. Moreover, an individual wire conduit for each motor is placed in each half of the robot body. As a result, the biopsy payload is assembled in each half separately before the clamshell pieces are fastened together, which substantially reduces assembly time and complexity.

Finally, the lead screw housing, collar pathways, and linkage pathways are all finished with wet sanding to reduce friction during actuation. The lead screw is rounded to match the outer face of the cylindrical robot housing. This eliminates stress concentrations and sharp edges that can damage the robot housing during actuation. Also, the collar and linkage pathways fit the mechanism components so that there is no binding or rocking during actuation. A schematic of the biopsy grasper mechanism is shown in Figure 3-7.

Figure 3-7: Schematic of Biopsy Grasper Mechanism
3.5 Flexible Biopsy Grasper

The designs in the previous sections have all involved a rigid biopsy grasper. The modular platform with the rigid biopsy grasper payload can be inserted into the abdominal cavity through a small incision. However, the proposed mechanism does not fit through a standard circular laparoscopic trocar, and therefore, the design must be refined. These refinements are shown in a schematic of the flexible biopsy grasper arm in Figure 3-8. When fully implemented, the grasper is affixed to the distal end of an arm that is typically perpendicular to the body of the modular robot platform. To allow the robot and grasper to be inserted and retracted through a trocar, the collar is split into two pieces and the anchoring method is modified. These components are connected together with a unique support structure that provides rigidity during sampling and the ability to flex the grasper arm 90º for insertion and retraction through a circular trocar.

The two collar pieces are connected to one another by two 0.30 mm thick Nitinol ribbons that are anchored to the walls of the inner and outer collars. A third Nitinol ribbon, connected to the biopsy grasper on one end and to the robot body on the other, anchors the grasper when the collars were actuated. This ribbon is profiled for clearance around the lead screw linkage.

Initially, due to the limited thickness of the collar walls, a Nitinol wire was used to anchor the bottom jaw to the robot. However, the wire is not rigid enough, and as a result easily bowed and coiled. On the other hand, thin ribbons can be flexed easily and will spring back straight without kinking, physical deformation, or memory effects. During insertion, the grasper arm can be flexed 90º. Once through the trocar, the ribbons will
return the grasper back to their normal orientation. A prototype of the flexible biopsy grasper is shown in Figure 3-9.

**Figure 3-8:** Schematic of Flexible Biopsy Grasper in the Normal (top) and Flexed (bottom) Configurations (© ASME 2008)

**Figure 3-9:** Flexible Biopsy Grasper Prototype
Because the only forces applied to the grasper arm during actuation are along the axial direction of the arm, the Nitinol ribbons will not flex during actuation. Furthermore, organs and tissue within the peritoneal cavity do not have sufficient rigidity to flex the Nitinol ribbons as the robot maneuvers to various sampling sites. Consequently, the grasper arm will remain perpendicular to the robot body during sampling and robot navigation. Moreover, the Nitinol ribbons restrict any flexion in the direction of wheel rotation, maintaining the rigidity needed by a tail during robot navigation. Finally, during retraction, the grasper arm will again flex 90º as it comes into contact with edge of the trocar.

Other refinements were made to the linkage and the inner half of the collar. The rigid biopsy grasper uses two linkages; each pinned on one side of the lead screw nut and the collar. However, the flexible design features a single linkage that is forked on one end to attach to the lead screw nut, and the linkage is pinned to the collar on the opposite end. Specifically, the linkage is pinned to the inner surface of the collar, and the combined length of the inner and outer collar is such that it is longer than the stroke length of the lead screw. This ensures that the outside surface of either the inner or outer collar is always in contact with the opening in the robot housing so as to minimize fluid infiltration while the robot is in the \textit{in vivo} environment.

Unfortunately, the single linkage tends to tilt or rock during actuation. While the biopsy grasper is still able to close completely, the energy and friction losses due to this rocking motion reduce the efficiency and reliability of this mechanism. Consequently, an alternative flexible biopsy grasper was designed and is shown in Figure 3-10.
For this design, the flexion of the biopsy grasper is a result of using a tension spring to connect the bottom jaw to the robot housing, and the spring fits within the collar. The rocking motion is reduced by using two linkages, one on each side, which is similar to the rigid biopsy grasper design. As the collar moves outward over the bottom jaw, the spring is exposed and the jaw was allowed to flex. On the other hand, when the collar moves inward, the flexion of the spring is eliminated as the collar comes into contact with the robot body. However, a longer collar and bottom jaw are necessary to implement this design so that the biopsy grasper would be rigid during actuation.

![Figure 3-10: Flexible Biopsy Grasper with Spring](image)

### 3.6 Actuation Forces

Biopsy graspers based on variations of the preliminary rigid design were constructed and incorporated into a bench-top jig for more extensive testing. The principal goal was to investigate the effects of different grasper profiles and lengths on the forces required to actuate the mechanism. Likewise, the maximum forces that could be applied by the mechanism were also measured. The test jig includes a load cell that is used to measure
the tensile force in the nylon supporting rod when the collar is actuated. Figure 3-11 shows the test jig with the motor, linkage, lead nut, collar, biopsy grasper, and load cell. For these tests, the nylon rod extended out of the distal end of the bottom jaw and was threaded into a Delrin connector attached to the load cell. The support brackets shown were used to keep the nylon rod from bending during actuation, and ensure forces transmit mainly in a longitudinal direction.

Measurements of actuation forces were made for graspers with a wide range of jaw lengths, opening angles, top jaw profiles, and collar geometries. Required actuation forces were determined by using the motor and lead screw linkage to slide the grasper collar over the jaws until closed. For each actuation, the required force was recorded starting with the grasper completely open and continuing until the grasper was closed. Maximum actuation forces were determined by recording the forces that could be applied with the collar held fixed at various positions corresponding to different times during the actuation process. Each complete test consisted of 50 separate actuations of the biopsy grasper. Load cell readings were recorded during each actuation at a rate of 20 Hz.
The length of the collar was varied to determine what effect collar length may have on the forces required to close the grasper. While the length of the collar appeared to have a minimal effect on the required actuation force, experiments concluded that a longer collar is less likely to bind and cause areas of high friction. Experiments also suggested that if the linkage attached to the collar near its center of mass, the collar was also less likely to bind during actuation. Next, a chamfer was implemented at the end of the collar. This also had a limited effect on the required actuation force. Finally, the material of the collar was varied. Observations of the mechanism suggest that a stainless steel collar is less likely to bind or stick than a plastic collar, resulting in a smooth and consistent actuation.

In addition to angled top jaw profiles, curved top jaw profiles were also tested using the jig. However, when compared to the angled profiles, the curved profiles required nearly three times the amount of force to close the grasper. Experimental results suggest that sharp bends in the top jaw profile are easier to flex than the curved profiles.

Finally, experiments using the test jig confirmed the orientation of the actuation mechanism. Initially the test jig was set up so that the lead screw mechanism pulled the collar over the top jaw of the grasper. Next, the jig was rearranged in the orientation that is shown in Figure 3-11, where the collar was pushed over the top jaw of the grasper. This allowed the greatest mechanical advantage to occur when the grasper was closed and the grasper would begin to penetrate the tissue sample. One can think of this push versus pull as a kinematic inversion that reverses the effect of the singularity. Moreover, the test results indicated that the required actuation forces for the “pulling” orientation (shown in Figure 3-12) were nearly twice as high as the required actuation forces for the
“pushing” orientation (shown in Figure 3-13). Observations during these tests showed that as the grasper closed in the ‘pulling’ orientation, the loss of mechanical advantage along the axis results in large friction forces due to binding between flexed top jaw and the collar.

Figure 3-12: Required Actuation Forces for “Pulling” Orientation

Figure 3-13: Required Actuation Forces for “Pushing” Orientation
The results of these tests were used to develop a final candidate for the top jaw profile. This candidate is approximately 12 mm long, and has an opening angle of 23°. Mean results from the required force test for this grasper are shown in Figure 3-14. The error bars indicate the standard deviation in the measured forces at intervals of approximately 1.8 seconds. The maximum required actuation force of 2.83 N was at the very start of the motion of the collar due to the need to overcome static friction and to begin flexing the top jaw of the grasper. The force decreased with time as the contact point between the collar and the top jaw moved farther away from the anchor point. The test results indicated that a maximum force of approximately 3 N was required to close the biopsy grasper.

![Figure 3-14: Required Actuation Forces for Rigid Biopsy Grasper (© IEEE 2009)](image-url)
Following the required actuation force tests, the jig was reconfigured to measure the maximum force that can be applied by the motor with the collar fixed at various positions corresponding to the grasper jaws ranging between fully open and fully closed. The motor was operated using a power supply with the same voltage (4 V) as the battery uses to power the robot motors. The maximum stall current was approximately 400 mA, well below the peak current (2.5 A) that can be supplied by the battery. At the start of actuation, with the biopsy grasper fully open, the maximum force that can be applied is 7.3 N. This is greater than twice the force required to begin closing the jaw. The angle between the collar and the lead screw linkage decreases as the collar translates along the biopsy arm. As a result, the amount of force that can be applied increases. Furthermore, a maximum applied force of approximately 13.2 N is attained at the translation limit of the collar at which point the grasper is completely closed. This is approximately 5 times larger than the required closing force.

These tests were conducted in a manner to overestimate the forces actually needed for the final biopsy grasper design. For example, a plastic collar was used during the tests, and friction forces are reduced if a stainless steel collar is used. Also, the Nitinol ribbon used for the top jaw was twice the thickness of the Nitinol ribbon used in the final grasper design. The overestimation of the experiments is meant to compensate for the unknown forces involved in grasping living liver tissue.

3.7 Modular Biopsy Grasper Prototype

The bench-top experiments identified and characterized the general mechanical and geometric design of a robotic biopsy grasper mechanism. The biopsy grasper prototype is assembled within the payload half of the wireless wheeled modular robotic platform.
(shown in Figure 3-15). The overall length of the robot body diameter plus grasper arm is greater than the diameter of most laparoscopic trocars. Therefore, during initial *in vivo* tests, the robot was inserted through a small incision. While a flexible arm was one of the early design goals, the immediate focus is on a small rigid arm (shown in Figure 3-16) that can successfully biopsy liver tissue.

![Figure 3-15: Modular Robotic Platform with Biopsy Grasper (© IEEE 2009)](image)

After successfully constructing a biopsy grasper prototype, the design and development of other payload options was explored. The implementation of different surgical tools into the same robotic platform demonstrates the modular design characteristics of these devices. The modular design of the robotic platform allows these different robot variations to be created quickly because only a few components differ from those in the biopsy grasper prototype.
Chapter 4

4 Other Payload Variations

4.1 Physiological Sensor Package

During laparoscopic surgery the abdominal cavity is insufflated with CO$_2$ to provide maneuvering space for tools and instruments. The temperature, pressure, and (sometimes) the humidity of this gas are monitored only externally to the body, and local conditions can vary dramatically. It is important for patient health and well-being during surgery that stable conditions are maintained. Without local measurements of these parameters, the actual conditions can only be estimated.

The schematic in Figure 4-1 illustrates the modular robot platform with a physiological sensor package in the payload area. Careful attention was paid to the layout and orientation of the sensor package. A ribbon cable connects the sensor package to the main control board. If the sensor package is not oriented properly, the ribbon would act as a spring when the clamshell pieces are fastened together. These
unnecessary forces would make assembly more difficult and cause the clamshell pieces to split apart slightly.

![Schematic of Modular Robot with Physiological Sensor Package](image)

*Figure 4-1: Schematic of Modular Robot with Physiological Sensor Package*

A flexible spring, similar to the one used in the design concept of the flexible biopsy grasper, is glued to the robot housing to serve as the tail. The payload area of the modular robot platform contains a specific space for the sensor package. Once again, the PolyJet manufacturing techniques allow customization of the robot body so that the circuit board fits securely, reducing the risk of wires disconnecting while the robot operates.

A prototype equipped with the physiological sensor package is shown in Figure 4-2. This custom designed circuit board is configured to monitor the temperature (T), pressure (P), and relative humidity (RH) within the abdominal cavity. The module also includes additional electrical components and circuitry for power conditioning, power management, etc. This module requires connections with the main control board for power and data communication only.

Temperature and relative humidity are measured using a single chip sensor module (Sensirion SHT15). This chip provides a calibrated digital output for both temperature and relative humidity via an on-board 14-bit analog-to-digital converter. The data are transferred to the master MCU via a 2-wire serial interface.
Figure 4-2: Exploded View of Modular Robot with Sensor Package (© IEEE 2009)

Pressure is monitored using a Freescale Semiconductor absolute pressure sensor (MPXH6300A). This sensor has a full range of 300 kPa, and a sensitivity of 16.2 mV/kPa. A regulated charge pump (Microchip MCP1252) is used to boost the 3.0 V supplied by the master control board to the 5 V required by this sensor. Integrated on-chip conditioning networks provide a high output, temperature compensated signal. This signal is measured by the master MCU using its analog-to-digital converter.

This platform is used to demonstrate the feasibility of in vivo sensory feedback. However, a variety of other sensors (e.g., pH, glucose level) can also be accommodated. For example, acidity levels (pH) within the peritoneal cavity can alert the surgeon to problems that may be harmful to the patient. For instance, a small tear or cut in the bowel may occur during surgery. While such perforations can be difficult to detect visually, significant relative changes in acidity levels can be used as a marker to help avoid surgical complications.
Tissue oxygenation is also a useful physiologic property. For example, measuring the oxygenation of blood can detect the onset of ischemia (insufficient blood flow) before it causes surgical complications [52]. Likewise, tactile sensors can detect local mechanical properties of tissue. Compliance, viscosity, and surface texture are all indications of the health of the tissue [6]. Sensory feedback from within the abdominal cavity can help generate a map of physiological properties over the surface of the abdominal organs, allowing the modular robotic platform to perform diagnostic surgical assistance.

4.2 Staple Payload

Another payload variation incorporates a stapling tool in place of the biopsy grasper. This stapling arm is designed to hold and close a common laparoscopic surgical staple. In this design, the collar is actuated in the direction A in the same manner as the biopsy grasper mechanism. Likewise, in the stapling design, the collar presses the top jaw against the bottom jaw in the direction B to close the surgical staple, as illustrated in Figure 4-3.

![Figure 4-3: Schematic of Stapling Arm Payload (© IEEE 2009)](image)

A prototype of the stapling payload is shown in Figure 4-4. The profile of the top jaw fits the shape of the staple, and plastic pads are used to hold the staple in the jaws. These
pads are fixed to the jaws, and they contain a channel for the staple to snap into place. Once snapped into place, the staple is securely held within the jaws until the collar is actuated. Other than the arm, the rest of the mobile robot is unchanged from the modular robot with biopsy grasper discussed in Chapter 3.

![Figure 4-4: Stapling Payload Prototype with Staple Unloaded (left) and Loaded (right) (© IEEE 2009)](image)

A traditional laparoscopic stapling tool applies pressure to the back edges (near the ‘U’) of the staple allowing the staple to collapse forward as it smashes shut. Bench-top experiments indicate that this arm design will only close the open end of the staple during actuation. Unfortunately, if the entire staple does not close together, the stapling arm is unable to effectively staple things close. Before further experiments are carried out, a new staple design must be tested. For instance, a ‘V’-shaped staple that matches the top jaw profile used for the biopsy grasper will close more uniformly. The current design is unable to accomplish this because the tip of the top jaw comes into contact with the bottom jaw before the entire staple is smashed together. However, if the top jaw profile is simply an upward angle and a staple is designed to fit that profile, the entire staple will smash together completely as the collar slides over the top jaw.

The stapling payload is simply a demonstration of the different tools that are possible with this actuation mechanism. For immediate surgical task assistance, the stapling
payload is not a feasible option for this mobile platform. To illustrate, current laparoscopic stapling tools have a magazine of staples so that several can be delivered without removing the tool from the surgical field. This payload design only facilitates one staple at a time. Therefore, this grasper design is more feasible for tissue manipulation or handling.

4.3 Clamping Payload

Another payload option similar to the biopsy and stapling graspers is designed to clamp blood vessels with enough pressure to stop bleeding. For this application the top jaw and bottom jaw are fitted with plastic inserts that are designed to be smashed together. As the collar actuates over the top jaw, the two pads fit together tightly, providing an area of high clamping force capable of stopping a bleeding blood vessel.

Preliminary bench-top experiments of a clamping device are simple tests to determine whether or not enough jaw force could be produced to stop blood flow. For the prototype, the blades are removed from the jaws of the biopsy grasper. Next, plastic inserts are formed to fit together tightly. The first experiment involved surgical tubing, and unfortunately, the tubing is too thick and stiff for the clamping jaws to squeeze shut. It is obvious that the surgical tubing is much thicker and stiffer than blood vessels. Therefore, the next experiment uses heat-shrink tubing to represent the blood vessel.

A small water pump is used to pump water through the heat shrink at a steady rate. The heat-shrink tubing is placed in the jaws of the clamping payload. In order to conserve the use of batteries for these experiments, a power supply is used to power the motor for actuation. The clamping mechanism actuates many times, and each time the
water flow reduces to a very slow drip. The experimental setup is shown in Figure 4-5, and a view of the clamping jaws that are tested, shown in Figure 4-6.

![Image of setup](image)

**Figure 4-5:** Setup of Bench-Top Experiments for Clamping Payload

![Image of clamping jaws](image)

**Figure 4-6:** Clamping Jaws Tested During Bench-Top Experiments

The clamping jaws used in the bench-top experiments are not optimally designed to stop blood flowing through a vessel. These jaws are adapted from prototypes damaged during the testing and development of the biopsy grasper. Small plastic inserts designed specifically to stop blood flow can easily be manufactured using the PolyJet techniques. These specifically designed clamping inserts are smooth and rounded in order to reduce vascular damage. Figure 4-7 shows a schematic of a redesigned clamping jaw that is designed to stop blood flow.
Traditional clamping devices, such as the mosquito clamp, are commonly known to cause vascular damage. The ribbed surface of the mosquito clamp causes areas of stress concentrations that can not be identified in a reliable way. Famaey et al. researched loading effects of mosquito clamps, showing that loads greater than 0.5 N can cause some vascular damage. However, it has been shown that, clamping at much higher loads with a smooth surface will cause less damage to the elastic lamellae and no visible inflammation reaction, when compared to a mosquito clamp, but the functional results were similar [53]. Their findings were based on immediate evaluation after tissue loading. In reality, the tissue will heal during the recovery time. Moreover, the findings suggest that patient trauma and recovery times are reduced when using smooth clamps compared to mosquito clamps.

Famaey et al. suggest that more experiments with different clamping durations, different speed of clamping, and different periods of recovery should be performed to help evaluate the effect that clamping for long periods time has on patient recovery time. The robotic platform discussed in this thesis could be used as a tool to help facilitate these experiments. For example, speed, force, and duration can all be controlled accurately, and the experiments could be performed *in vivo*.
4.4 **Payload Compatibility**

The biopsy, stapling, and clamping payloads all rely on the exact same actuation mechanism. All of these grasper arms are anchored to the modular robot through a mounting hole opposite the grasper arm. The nylon insert that the top and bottom jaws are glued to is made to tightly fit this mounting hole. The arm must be twisted back and forth into place, and static friction was sufficient to keep the arm in place during collar actuation.

As a result, it is unnecessary for the grasper arms to be glued to the robot housing. Consequently, the compatibility between grasper arms became another modular feature of the robot design. The biopsy, stapling, and clamping arms, as well as any other designs that use the same actuation mechanism, can be exchanged and ready for insertion within a few minutes. One can imagine a tool belt of various end effectors ready to equip the modular wireless wheeled robot platform for whatever specific surgical assistance is required.

4.5 **Cautery Payload**

The dissection of tissue, arteries, and other anatomical features such as the bile duct is common during laparoscopic surgery. While previously discussed payload variations focus on the sampling and manipulation of tissue, these designs are impractical when it comes to dissecting tissue. Therefore, a disposable cautery device, commonly used during surgical procedures, is implemented into the payload area of the modular robot platform.

One of the first design concerns of the cautery payload was how the PolyJet material would hold up to the intense heat of the cautery tip. Therefore, laboratory experiments
were conducted that placed a disposable “pen” cautery tip in contact with the robot housing material. Results showed that only the very tip of the cautery tool caused damage to the PolyJet material. If any other part of the cautery is in contact with the robot housing, the heat is not sufficient to damage the material. Because the very tip of the cautery tool should never be in contact with the robot housing, there are no concerns that using a cautery payload will damage the modular robot housing.

While the heat of the cautery tip did not damage the robot housing, it did produce another design concern. For example, when the cautery tip was activated for long periods of time, the solder used to connect the cautery tip pads to the wires melted, and the connection failed. A second attempt was tested, and with this attempt, the cautery pads and wires were covered in ultraviolet cure epoxy in order to hold them in place during use. This technique was successful as the epoxy served as a casing to hold the solder in place during long periods of cautery testing.

The thin wires in the tip of the cautery are very fragile and easy to bend. While organs and tissue inside the abdominal cavity do not have sufficient stiffness to damage the cautery tip, there were concerns that the tip could be damaged during insertion or retraction. Therefore, unlike other payload end effectors, the cautery tip is designed to completely retract into the robot housing, as shown in Figure 4-8, so that the tip is not exposed during robot insertion, navigation, and retraction.
It is important to stay within the concept of modular design and interchanging robot components. Therefore, the same lead screw and motor used in the biopsy grasper design are used to actuate the inward and outward movement of the cautery arm. In order for the cautery arm to entirely fit inside the robot payload area during navigation, the cautery tip needs to mount to something small. A novel solution to this problem fixes the cautery tip to a 0.25 mm thick Nitinol ribbon. This is the same type of ribbon used in the biopsy grasper top jaw. A small plastic pad is designed to fit the cautery tip and provide insulation between the wire cautery tip and the Nitinol ribbon. A hole is drilled through the ribbon on the opposite end, and the ribbon is fastened directly to the lead screw nut.

The Nitinol ribbon is curved using the same ‘shape setting’ heat treatments described in Chapter 3. A stainless steel pin shown in Figure 4-9 is used to flex the ribbon outward as the lead screw actuates. The rigidity of the pin is necessary because the sharp corners of the thin Nitinol ribbon can damage the soft PolyJet material during actuation. When the cautery arm is fully extended, the lead screw and steel pin provide adequate pressure to the ribbon so that the cautery tip is rigid and stable during operation.
The cautery payload requires a large power source in order to provide the current necessary to heat the cautery tip. A handheld pen cautery uses two AA alkaline batteries to power the cautery. During the prototype stages, the simple solution is to add a battery pack, as shown in Figure 4-10. This eliminates any redesign of the modular portion of the robotic platform. A relay circuit, shown in Figure 4-11, connects the battery pack and cautery switch to the main control board. The relay circuit is housed inside the battery pack, and a tether runs from the battery pack to the mobile robot platform.

![Figure 4-10: Schematic of Cautery Battery Pack](image1)

![Figure 4-11: Cautery Relay Circuit Board](image2)
A modular robot prototype equipped with a cautery payload was used in a series of laboratory demonstrations. One of these experiments used the cautery arm to cut a rubber band. A sequence of video frames shown in Figure 4-12 illustrates the experiment. First the modular platform was driven near the rubber band with the cautery arm retracted within the robot housing. After the cautery arm was extended, the cautery tip was activated. Next, the glowing hot cautery tip was applied to the rubber band, and the rubber band was dissected. Finally, the cautery arm was retracted so that the robot could navigate to another area.

A final bench top experiment, shown in Figure 4-13, involved the cooperation of two robot platforms. One was equipped with a clamping payload, and the other was equipped
with a cautery arm. The original positions are shown in Frame 1. The objective of the experiment was for the clamping robot to grasp the rubber band at the region just above the green marker. Next, the rubber band was stretched tight, and the cautery robot moved in to cut the rubber band in Frame 6. Then, the clamping robot repositioned the dissected rubber band, and the cautery robot moved in for another dissection in Frame 9. Finally, the two pieces of rubber bands were released, and the robots were returned to their original position.

Figure 4-13: Bench Top Testing of Cooperative Stretch and Dissect

4.6 Ceiling Pan/Tilt (CPT) Payload

The modular robot platform can also provide vision assistance in an effort to replace the laparoscope. Unlike the payload variations discussed previously, the ceiling pan/tilt (CPT) payload requires an additional component to be added to the modular robot
platform. Specifically, an outer housing that can be magnetically attached to the inside of the abdominal wall is created by making a small modification to the basic robot design, as illustrated in Figure 4-14. The payload area contains a color complementary metal oxide semiconductor (CMOS) imager and white light emitting diodes (LED) that can be used to provide visual feedback from the operating field.

![Figure 4-14: Schematic of CPT Payload Configuration (© IEEE 2009)](image)

The modular design for the mobile robot platform is modified by replacing the two wheels with a single outer cylindrical housing supported by bearings at both ends of the modular robot. This outer housing also incorporates a viewing slot for the camera and LEDs, and a magnet at each end for attachment to an external magnetic handle. One end of the outer housing serves as a cap so that the clamshell pieces of the modular platform can fit entirely inside. The magnetic attachment allows a surgeon to move (pan) the in vivo camera by sliding the external handle across the patient’s abdominal wall. A single PMDC motor and spur gear combination in the robot allows the inner robot housing body containing the camera payload to rotate (tilt) relative to the outer housing.
The camera system developed for use in the CPT, shown in Figure 4-15, includes an MT9V125 color digital CMOS image sensor from Micron. This ¼-inch sensor has a 640x480 active pixel array with both NTSC and PAL analog composite video outputs. It also has numerous features that could be used to optimize image quality, such as color correction, white and black balance, and gamma correction. A matched, multi-element Sunex lens (DSL758C) is used in a fixed-focus configuration that provides a depth of field of 4 cm to infinity. Illumination is provided by 5 mm diameter 20,000 mcd white LEDs (LED5 15DG). A wider illumination viewing angle (and more diffuse illumination pattern) is obtained by carefully filing off the LED lens assembly from the body of the device. The video output from the CPT prototype, shown in Figure 4-16, is transmitted to the operating room monitor using a small tether because a miniature wireless video transmitter providing adequate video quality was unavailable. All power is provided by the on-board battery, and command and control data transmission is wireless.

Figure 4-15: Camera Circuit Board for CPT Payload (© IEEE 2009)

Figure 4-16: Prototype of CPT Payload (© IEEE 2009)
Zooming capabilities are extremely important for many surgical tasks. While many traditional laparoscopes do not have a zoom mechanism, a surgeon can adjust the image by moving the laparoscope in and out of the trocar. It is possible for an adjustable focus and/or zoom mechanism to be added to the payload area of the robot with some miniaturization of the camera system circuitry. This would allow enhanced vision assistance without having to move the mobile platform closer to the surgical field.

This concludes the discussion on a large variety of surgical tools for vision and task assistance that are implemented into the modular robotic platform. Biopsy grasper, staple, clamping, cautery, and vision prototypes were produced and tested during laboratory demonstrations. In many of the discussed payload variations, the size of the payload area can accommodate a combination of options. For instance, both the sensory package and mechanical linkage payload options have enough room to include a camera system. As a result, it is feasible for the operator to have visual feedback from many of the suggested payload variations. The remaining sections of this thesis discuss an analysis of the actuation mechanism and biopsy grasper design, as well as the in vivo test results of these payload variations.
Chapter 5

5 Biopsy Grasper Design Parameter Analysis

5.1 Model of Biopsy Grasper

A model of the biopsy grasper discussed in Chapter 3 is created using the commercial finite element analysis software called Abaqus. The primary goal of this analysis is to evaluate the design characteristics of the biopsy grasper. For example, the mechanical efficiency of the collar actuation as a function of Nitinol ribbon angle and thickness is investigated. Also, the forces that are translated to the cutting edge of the biopsy grasper as a result of required actuation forces are investigated.

In order to investigate the design characteristics, an approximate model of the biopsy grasper must be created using Abaqus. This model must be validated and compared with the experimental results discussed in Section 3.6. Also, model components are the same size as their corresponding components discussed in Section 3.6. The initial simulation is set up using the biopsy collar and bottom jaw.
The material used for the collar, bottom jaw, and blades is stainless steel. The stainless steel material properties used in the simulation have a Young’s modulus of 193 GPa, a Poisson’s ratio of 0.3, and a yield stress of 205 MPa. The pads used to mount the top and bottom blades of the biopsy grasper are also included in the simulation. The PolyJet material characteristics have a Young’s modulus of 2.87 GPa, a Poisson’s ratio of 0.35, and a yield stress of 110 MPa. Figure 5-1 shows a schematic of the model before actuation begins.

The top jaw of the biopsy grasper is also included in Figure 5-1. However, other mechanical features such as pins, fasteners, and nylon inserts are left out to reduce the complexity and computational costs of the simulation. The material properties for the Nitinol ribbon top jaw of the biopsy grasper are a Young’s modulus of 83 GPa, a Poisson’s ratio of 0.33, and a yield stress of 690 MPa. These are the average property values of austenitic Nitinol. Phase diagrams suggest that 700° Celsius is required for a phase transformation to martensite. However, the shape-setting heat treatment process only uses temperatures of 500° Celsius. Finally, an initial test is performed to ensure initial simulation and boundary conditions are interacting as expected.

Figure 5-1: Model Schematic of Biopsy Grasper
A finite element mesh is approximated for both the collar and top jaw by applying well defined loads and indentifying converging results. A very fine mesh size is needed to approximately model the contact interaction between the collar and the ribbon. If the mesh size is not small enough, the slave surfaces and master surfaces would intersect and overlap. The mesh size for the collar and ribbon are 0.4 mm and 0.09 mm respectively is shown in Figure 5-2. Both the ribbon and collar mesh use a C3D8R element, which is a linear, 8-node, rectangular element with reduced integration.

![Figure 5-2: Model Schematic of Collar and Ribbon](image)

A displacement boundary condition, which simulated the translation of the collar during actuation, is placed on the collar to control its motion during the simulated actuation. The collar displacement boundary condition is limited so that both sides of the collar move together in unison. This restriction simulates a uniform translation the two linkages, which does not allow the collar to twist during actuation. A second boundary condition is placed on the Nitinol ribbon so that the glued end of the ribbon is fixed during actuation. This boundary condition simulates the nylon insert that is glued inside the bottom jaw. The ribbon is also glued to this nylon insert so that it does not translate in the axial direction during the collar actuation. As a result, only the Nitinol ribbon can flex during actuation.
Careful attention must be paid to the interaction between the stainless steel collar and the Nitinol ribbon that forms the top jaw of the biopsy grasper. More specifically, the contact interaction between the collar and the ribbon is very complicated because only a small amount of the ribbon is actually in contact with the collar. This is a result of the thin flat ribbon and the circular surface of the collar.

Figure 5-2 showed the contrast between the mesh sizes of the collar and the ribbon. The interaction is modeled using a master-slave relationship. To illustrate, the master surface (collar) has a much coarser mesh than the slave surface (ribbon). This relationship ensures that nodes of the slave elements are able to deform around the nodes of the master elements. A “hard” kinematic contact interaction property is used for any interactions occurring normal to surfaces in the model. Any tangential interactions are modeled with friction coefficients, which are determined from literature sources. The friction coefficient for the interaction between the stainless steel collar and the stainless steel bottom jaw is 0.42. Also, the friction coefficient for the interaction between the Nitinol ribbon and collar is 0.64. Both interactions are modeled as surface to surface interactions so that the small contact areas of the ribbon edges are well represented.

Images from the simulation (Figures 5-3 through 5-5) show the stress distributions on different components of the biopsy grasper. For instance, the biopsy collar is shown in Figure 5-3. The areas of stress concentrations are near the fastener holes, which are a result of the displacement boundary conditions applied at those locations. It is also clear to see the complicated interaction between the top and bottom jaws of the biopsy grasper and the collar. Small areas of stress concentrations form as the edges of the jaws come
into contact with the collar during the simulated actuation. Units in Figures 5-3 through 5-5 are in MPa.

**Figure 5-3:** Stress Distribution on Collar When Grasper Is Closed

As the collar actuated, the ribbon flexes and the top blade is pressed against the bottom jaw. The stress distribution on the top blade is shown in Figure 5-4. The two high stress concentrations on the top of the blade are located where the edges of the Nitinol ribbon come into contact with the top pad and blade.

**Figure 5-4:** Stress Distribution on Top Blade during Actuation
Finally, different images were taken to show the animation of the simulation. In Figure 5-5, the left-hand images show the entire biopsy grasper assembly during
actuation. Meanwhile, the right-hand images show the same assembly with the biopsy collar removed so that the deflection and stress distribution of the Nitinol ribbon can be seen more clearly. The images show the initial position of the collar before actuation began in the top images. The simulation progresses until the biopsy grasper is completely closed and the stall force is reached by the collar, which is shown in the bottom images.

The stress distributions of the model indicate the important areas of interaction between model components. These areas are investigated to ensure that the model adequately represents observations from laboratory experiments. Furthermore, by analyzing these interactions, a valid finite element model of the biopsy grasper is created in order to perform simulated tests without constructing different prototypes.

5.2 Validating the Model

It is important to verify that the finite element model constructed using Abaqus is similar to the actual biopsy grasper device. For example, the experimental data described in Section 3.6 are compared to the force readings calculated in the simulation. The displacement boundary condition is applied to four nodes on the collar, located on the inner and outer walls of the fastener mounting holes. The reaction forces at these four nodes are summed to represent the total force needed to actuate the collar and close the biopsy grasper.

The simulation results indicate that a force of 4.28 N is required to close the biopsy grasper. By comparison, the lab experiments only required forces with an average of 2.83 N. However, Nitinol is a super-elastic material, and its stress versus strain responses differ in tension and compression. These differences, along with the material
composition, lead to the large range of possible material property values. There is some error in the model because the Nitinol is modeled as a purely elastic material.

The first simulation uses a yield stress of 690 MPa, which is at the maximum of the suggested range of 195-690 MPa. Table 5-1 below shows the results of the force necessary to close the grasper with various yield stress values for the Nitinol ribbon. The table shows that a yield stress of 210 MPa results in a force of 2.85 N to close the biopsy grasper. This theoretical force is almost exactly what was measured during lab experiments. More discussion on the effects of the Nitinol material properties will follow.

![Figure 5-6: Required Actuation Force with Respect to Nitinol Yield Stress](image)

The calculated actuation force is plotted with respect to time and compared to the lab experiments discussed in Section 3.6. The comparison is shown in Figure 5-7. There is a small delay in the start of the simulation because unlike the lab experiments, the collar is
not initially in contact with the Nitinol ribbon. The rise time and peaks of the two plots are very similar. Also, both plots decrease as the collar continued to actuate to its final position and the biopsy grasper is closed. However, one noticeable difference between the simulation and experimental data is the slope of the decreasing line.

![Figure 5-7: Comparison between Model and Experimental Data](image)

Design parameter analysis of the biopsy grasper design requires a finite element model that resembles the experimental data as closely as possible. If there are discrepancies between the theoretical model and experimental data, then those differences must be investigated and thoroughly explained. With this particular model, there are several areas that could lead to the differences. For example, the effects of friction coefficients, collar length, and Nitinol material properties on the simulation are all investigated.
First, the effects of various friction coefficients are investigated. Simulation results indicate that varying the friction coefficients only shifts the force response left or right in the plot with respect to time. Moreover, the slopes, rise time, and decay rate of the force response are not changed by varying friction coefficients.

Next, the length of the biopsy collar is also varied. Several simulations indicate that the collar length does not significantly affect the force response during actuation. To illustrate, as the collar length varies, the starting point also varies to ensure that the simulation begins just before the collar is in contact with the Nitinol ribbon. However, the collar displacement remains the same because the ribbon profile does not change and requires the same amount of collar displacement to close the biopsy grasper.

Finally, the material properties of the Nitinol ribbon are varied in order to determine their effects on the force response of the simulation. The geometry of the top jaw ribbon creates a stress concentration at the fixed base of the ribbon. Careful attention is paid to the boundary conditions applied to the ribbon so that Nitinol strain limits are avoided. Experimental observations show no signs of plastic deformation or wear on the Nitinol ribbon. When the entire base of the ribbon is fixed solidly in the bottom jaw, a yield stress of 350 MPa is necessary to avoid reaching the strain limits of the Nitinol ribbon.

The experimental results measured a peak force of 2.83 N during actuation. That force declined steadily to approximately 2.25 N when the biopsy grasper was completely closed. Material property changes affect both the peak force and slope as the force decreases during actuation. To illustrate, if the Young’s modulus increases, the rise time, peak force, and the decay rate all increase. Likewise, if the yield stress increases, the rise time and peak force also increase. However, the decay rate tends to decrease as the yield
stress increases. Consequently, varying the material properties of the Nitinol does not effectively alter the simulation results in a manner to more closely resemble the experimental data, and therefore, the originally suggested material properties (Young’s Modulus of 83 GPa and yield stress of 690 MPa) are used for the remaining simulations.

After carefully examining video of the lab experiments, two areas needing further investigation are obvious. First, a thick Nitinol ribbon was used in the lab experiments. When the ribbon was too thick to be flexed during motor actuation, the ribbon was ground down to reduce the ribbon thickness to a value that could be flexed regularly during actuation. Second, the mounting slot that was machined into the nylon rod for anchoring the top jaw was not an exact tight fit. The clearance was filled with epoxy, and it was observed that the top jaw anchor has some flexion during actuation. Therefore, it is important to investigate what effects the ribbon thickness and flexion had on the force response of the collar.

Simulations are performed while varying the ribbon thickness from values of 0.26 mm to 0.34 mm. Figure 5-8 shows the force response of the biopsy collar during actuation. It is clear that the thickness of the Nitinol ribbon affects the rise time and magnitude of the force response. However, the decay rate of the force as the biopsy grasper is closed does not seem to be affected by the ribbon thickness. The yield stress used for Nitinol is 690 MPa during these simulations. While the ribbon thickness can vary due to the nature of the grinding process used to reduce the ribbon thickness, measurements and simulation results suggest that 0.30 mm is the most realistic approximation of the mean ribbon thickness of the top jaw.
Clearance in the mounting slot used to anchor the top jaw allows the ribbon to flex slightly as the biopsy collar comes into contact with the top jaw. Therefore, a second displacement boundary condition is applied near the bend in the Nitinol ribbon. This boundary condition allows the base of the ribbon to flex much like it does due to the clearance or ‘slop’ in the mounting slot. The model schematic of this flexion is shown in Figure 5-9. The arrow indicates the location that the flexion boundary condition is applied. Because the flexion occurs at the instant that the collar comes into contact with the top jaw, a step input is used for this displacement boundary condition.
The magnitude of this ribbon flexion displacement boundary condition is varied from 0.09 mm to 0.17 mm. The force response results of those simulations are shown in Figure 5-10. Once again, magnitude and rise time are affected by this flexion. However, one noticeable difference is that the amount of flexion seemed to directly affect the decay rate of the force response during actuation. An increase in flexion results in a decrease in the decay rate of the force response.

![Figure 5-10: Variation of Ribbon Flexion](image)

Finally, the best simulation is selected and compared to the experimental data. For this simulation, the ribbon thickness is 0.30 mm, and the flexion boundary condition is set at 0.17 mm. The force response, shown in Figure 5-11, calculates a peak force of 2.71 N. There is also a slight difference in the decay rate of the force during actuation. The magnitude differences can be attributed to slight differences in the frictional forces during
actuation. Meanwhile, while it is known that the ribbon base flexes during actuation, it is unknown whether the flexion is constant during the entire actuation. For example, flexion at the base of the ribbon may increase during actuation, and that results in a smaller decay rate. However, the properties of this simulation are sufficient to validate the model with the experimental results. Any differences between the model and the experimental measurements can be attributed to losses due to friction and mechanical inefficiency of the apparatus used to measure the force data in the lab experiments.

![Figure 5-11: Force Response of the Simulation](image)

5.3 Effect of Ribbon Thickness

Two different thicknesses of Nitinol ribbon are available for prototype construction of the biopsy grasper. The thicker ribbon (0.30 mm) was used during the lab experiments and early testing. However, using a thinner ribbon requires less force to actuate the collar. It is important that the biopsy grasper be as mechanically efficient as possible. The jaw force applied by the biopsy grasper needs to be maximized so that enough
pressure is present to cut and tear the liver tissue. Furthermore, if the thinner Nitinol ribbon (0.25 mm) can supply a similar amount of jaw force with less force needed for actuation, the biopsy grasper would be much more efficient.

Two different simulations compare the actuator force required to move the collar and the resulting jaw force. Once again a displacement boundary condition is used to move the collar. However, the collar is actuated until a stall force of approximately 8.5 N is reached in both simulations. This is done for comparison purposes and to more approximately represent the actual biopsy grasper. The motor actuates the prototype biopsy grasper until force supplied by the motor can no longer move the collar. To illustrate, when the top jaw comes into contact with the bottom jaw, the ribbon can no longer flex and the collar eventually came to a halt. Figure 5-12 shows the force response of the simulations, and Figure 5-13 shows a close-up of the force response during actuation. Both figures show that the thin ribbon requires less actuation force.

![Figure 5-12: Simulation Results of Different Ribbon Thicknesses](image)
Finally, the jaw force is compared between the two different ribbon thicknesses. Figure 5-14 shows that both the 0.25 mm and 0.30 mm Nitinol ribbon thickness results in a jaw force of approximately 3.25 N. However, the thin ribbon can supply the same jaw force as the thick ribbon with 40% less actuation force.
5.4 Effect of Top Jaw Profile Angle

Another major design parameter that can be investigated is the profile angle of the top jaw. The use of simulations to investigate the effects of the top jaw profile angle on the mechanical efficiency of the biopsy grasper is less expensive than constructing multiple prototypes. For these comparisons, the profile angle is varied between 21° and 25°. Once again, a displacement boundary condition is placed on the collar. For all of the various angles the collar is displaced until a stall force of approximately 8.75 N is reached.

![Figure 5-15: Force Response of Different Profile Angles](image)

During the simulations the jaw force and ribbon force are recorded. The ribbon force is the reaction force on the ribbon in the axial direction of the collar movement. The force response during collar actuation is shown in Figure 5-15. Likewise, a close up view during the actuation process is shown in Figure 5-16. Simulation results indicate that
increasing the profile angle increases the force needed to close the biopsy grasper and reduces the jaw force. These phenomena can be explained by likening the Nitinol ribbon to a spring. If the profile angle is increased, more displacement is necessary to close the biopsy jaws. Consequently, the Nitinol ribbon stores more energy, which results in less force translated to the jaws of the biopsy grasper.

![Figure 5-16: Actuation Forces of Different Profile Angles](image)

Finally, the calculated jaw forces resulting from different profile angles are shown in Figure 5-17. There is a distinct difference between the 21°, 22° and 23° profiles and the 24° and 25° profiles. This is a result of adjustments to the initial position of the collar that are needed in order to accommodate the different top jaw profiles. The 24° and 25° profiles are not ideal because the displacement necessary to close the biopsy grasper is larger than what is possible with the length of the power screw. The 21° and 22° profiles are nearly identical, and the 23° profile behaved similarly with only a small reduction in the jaw force produced.
Throughout the design parameter analysis, simulations are used to evaluate and calculate the forces and stress distributions on the different components of the biopsy grasper. Results indicate that a thin Nitinol ribbon reduces the actuation force needed to close the biopsy grasper, but the thin ribbon does not decrease the jaw force. On the other hand, the top jaw profile angle does affect the jaw force. As the angle increases, the jaw force decreases. However, if the profile angle is too small the opening of the biopsy grasper is unable to grasp onto the tissue easily. Therefore, a $23^\circ$ profile angle is identified as the optimal design. A $23^\circ$ angle utilizes the largest biopsy grasper opening without greatly affecting the jaw force and actuation kinematics.
Chapter 6

6 Background in Tissue Modeling

6.1 Introduction to Soft Tissue Modeling

Soft biological tissues have been researched extensively over the years in an effort to learn more about their material properties and their responses under loads. The computer graphics industry has been a large source of research involving the simulation of object deformation. New technology driven by the entertainment industry (computer animated movies) has allowed for special effects and entire movies to be rendered using these simulation techniques. Visual computer effects, which depend on complex simulation models, allow for very realistic simulations of actual events.

Another major area of research driving the modeling of soft tissues is the use of surgical simulation. Medical students must be trained to perform surgical operations. There are many issues that surround the training process. Animal models or human cadavers are two possible options used to train future surgeons. However, these methods
have ethical and economical constraints. Realistic surgical simulators could provide a desirable platform for surgical training.

This chapter provides a review of techniques used to create models of soft biological tissues. The purpose of this chapter is to summarize these techniques for the reader’s reference. First, a discussion on the characteristics of soft tissue helps to describe the complexity of the models needed for accurate simulation. Next, several assumptions and simulation methods are discussed. The pros and cons of each method are presented to evaluate the different approaches to constructing a soft tissue model. Finally, the specific application of cutting soft tissue is discussed. Cutting and manipulating tissue is a complex problem that needs to be addressed in order to develop an accurate simulation of surgical procedures. If the reader has no interest in a review of modeling techniques without affecting the understanding of the rest of this document, this chapter can be skipped.

### 6.2 Characterization of Liver Tissue

An important first step in developing a soft tissue deformation model is investigating the characteristics of soft tissue. The mechanical properties of soft tissues, such as the liver, are inexact and difficult to measure. This is due to the complex nature of the tissue itself. The relationship between stress and strain is not proportional in biological tissues [54]. Consequently, soft tissues are not purely elastic. Constant strain results in a stress relaxation. Constant stress results in creep. Soft biological tissues are anisotropic, due to the presence of reinforcing fibers in the extracellular matrix [55-57], and their stress-strain relationships are non-linear. These characteristics suggest that biological soft
tissues are viscoelastic materials. In general, soft tissue is compressible, non-homogeneous, anisotropic and viscoelastic [58].

Some basic simplifying assumptions are used to help reduce the complexity of soft tissue material properties. Very soft tissues are most often assumed to be incompressible under physiological pressure levels due to their high water content [59]. This assumption was validated experimentally during compression tests using brain tissue by observing that no fluid escaped during the tests. A purely incompressible material has a Poisson’s ratio of 0.5. Liver tissue is thought to be nearly incompressible. For example, Chui et al. concluded that the Poisson’s ratio for porcine liver in elongation and compression are 0.43±0.16 and 0.47±0.15 respectively [60].

Also, very soft tissues do not bear mechanical loads or exhibit directional structure, which leads to the assumption of the material to be isotropic. Finally, researchers often assume that the material is homogeneous. The homogeneous and isotropic assumptions are especially justified for large soft organs, such as liver and kidney. These organs have limited or no reinforcement by muscular fibers [61]. Many researchers begin formulating models by using these general assumptions. Once the models are evaluated, assumptions may be refined to provide a more accurate model.

One example demonstrating the complexity of biological tissue is the disparity present in current research and data regarding mechanical properties. The properties of tissues vary by a number of individual factors such as age, sex, clinical history and living habits [62]. Properties also vary depending on ex vivo, in vitro or in vivo experiments. For example, dead organ and muscle tissues typically stiffen over time [63]. However, in
vivo experiments have to deal with the affects of respiratory and circulatory motions that occur in living tissue [64].

Also contributing to data variation is the fact that many different techniques are used by researchers to estimate the exact values of various mechanical properties. For example, experimental measurements are performed on a wide variety of animals. Unfortunately, there is not sufficient data available to describe the similarities of liver tissue from different species (i.e., rabbit, porcine, human).

Modulus is the measure of an object’s resistance to deformation. For biological tissue, modulus is not constant, and it can vary from location to location. For simplification, many researchers use the case of small deformations to determine the elastic modulus of soft tissue. Biological tissues can be considered linearly elastic when deformations are less than 10%. However, recent research by Misra, Okamura, and Ramesh suggest that small deformations may even be described as less than 2% [65].

Samur et al. discussed approaches followed by researchers to measure soft organ modulus, which can be divided into two groups [63]. The first group used a hand-held instrument with position and force sensors. The second group used robotic devices to provide better-controlled stimuli. Modulus estimates based on the two approaches are dramatically different from one another, making it difficult to choose the material properties used in models. For example, some in vivo experiments estimated the elastic modulus of pig liver as 10-15 kPa [66]. Another robotic device used for in vitro experiments approximated the shear modulus of porcine liver at 40 kPa [61].

Hu et al. performed studies on soft tissues using large deformations and concluded that modulus varies with strain rate [58]. To illustrate, the modulus was higher at lower
strain rates compared to higher strain rates. This characteristic was a result of the viscoelastic properties of soft tissue. Specifically, Rentschler et al. measured the viscoelastic properties of liver tissue, such as stiffness and damping coefficients [47]. For example, the stiffness coefficient of bovine liver was approximately 350,000 N/m³, while the damping coefficient was approximately 4000 N·s/m³.

Internal organs, such as the liver, essentially consist of a functional vascularized internal part (parenchyma), and an external capsule (stroma) [67]. The capsule, or membrane, is a thin but tough fibrous connective framework of densely interwoven collagen fibers. Test results concluded that neglecting the capsule leads to a significant overestimation of the liver’s mechanical properties.

Furthermore, the effects of the membrane were negligible in bovine tissue samples. However, in living tissues the membrane effects are expected to be more significant. For bovine liver, Rentschler et al. suggest that the membrane tension is approximately 1.0L (N), where L is the length of contact [47]. This value was determined from physical observation under normal loads, and approximated by incorporating the liver stiffness.

### 6.3 Soft Tissue Deformation

Classical elasticity theory suggests that the restoring force (stress) in a body is a single-valued function of the deformation (strain) of the body [68]. Another important assumption is that the relationship between stress and strain is independent of the history of deformation. For example, an ideal spring stores potential energy during deformation and releases it entirely as it recovers to the initial shape. In contrast, a perfect (Newtonian) fluid stores no deformation energy, and consequently, exhibits no resilience.
For most soft tissues, this linear elastic deformation is only valid for small displacements. For larger displacements, especially those present in cutting tissue, more complex nonlinear models have been introduced. Another limitation of the linear model is that it is not invariant with respect to rotations [69]. The elastic energy increases when an object undergoes a rotation. Consequently, this leads to a variation of the volume.

Most biological soft tissues are characterized as viscoelastic materials. Viscoelasticity combines elastic and viscous (Newtonian fluid) behaviors. Unlike elastic deformation, the instantaneous deformation of a viscoelastic model is a function of the entire history of applied forces [68]. Similarly, the instantaneous restoring force is a function of the entire history of deformation. Also, the response under compression depends on the strain rate and is nonlinear [70].

The equations of motion for deformable bodies have been discussed in many articles regarding basic mechanics. These equations can be written in Lagrange’s form [71] as follows:

\[
\frac{\partial}{\partial t} \left( \mu \frac{\partial \mathbf{r}}{\partial t} \right) + \gamma \frac{\partial \mathbf{r}}{\partial t} + \delta \dot{\mathbf{e}}(\mathbf{r}) = \mathbf{f}(\mathbf{r}, t) ,
\]

(1)

where \( \mathbf{r}(a,t) \) is the position of the particle \( a \) at time \( t \), \( \mu(a) \) is the mass density of the body at \( a \), \( \gamma(a) \) is the damping density, and \( \mathbf{f}(\mathbf{r}, t) \) represents the net externally applied forces. The first term is the inertial force due to the distributed mass. The second term is the damping force due to dissipation. Finally, the third term is the elastic force due to the deformation (strain energy).

A description of applied forces is necessary to determine the behavior of a deformable body as a result of those forces. Terzopoulos et al. present derivations of gravitational
forces, connection (or spring) forces, viscous fluid forces near the surface, and collision forces [72]. Some important conclusions were also discussed. For example, elastic bodies should not self-intersect as they deform. This can be avoided by surrounding the surface with a self-repulsive collision force.

6.4 Mass-Spring Models

Mass-spring models are relatively easy to implement. They have been used for both static and dynamic computation. Mass-spring models are made up of a set of points linked by springs and dampers. In the dynamic mass-spring system, the equilibrium equation has the following form [73]:

\[
M \frac{\partial^2 X}{\partial t^2} + D \frac{\partial X}{\partial t} + KX = F(X) \quad ,
\]

where \( \frac{\partial^2 X}{\partial t^2} \) and \( \frac{\partial X}{\partial t} \) are the first and second derivatives of \( x \) with respect to time, \( M \) is the mass matrix, \( D \) the damping factor matrix, and \( K \) the stiffness matrix. \( F \) denotes the external forces.

Mass-spring models are easy to construct. The springs constrain the length between vertices. Therefore, the number of springs per vertex affects the global behavior of the system. For example, in an under-constrained system, several rest positions are possible and the system can fall into unwanted local minima [74]. Conversely, if the system is over-constrained, the range of deformation is decreased.

Paloc discussed the advantages of tetrahedral mesh representation [75]. For example, tetrahedral meshes can accommodate Cartesian, rectilinear and curvilinear meshes. A tetrahedral mesh produces a more accurate representation of objects with relatively few
elements. Also, the mesh gives triangular surface representation, which is standard for most visualization software.

Another advantage of mass-spring models is their ability to handle both large displacements and large deformations [76]. The deformation of soft tissue can be simulated as a process of force propagation among the mass points on a nodal basis. For linear elastic springs and velocity-dependent dampers, the dynamics of a node $i$ can be formulated using Newton’s Law of Motion [77]. The nodal displacement of the $i$th node $u_i$ due to an external force $F_i$, is given by

$$m_i \ddot{u}_i + c_i \dot{u}_i + \sum_j k_{ij} \frac{r_{ij}}{r_{ij}} (l_{ij} - r_{ij}) = F_i,$$  \hspace{1cm} (3)

where $m_i$ and $c_i$ are, respectively, the mass and damping constant of the node $i$, $r_{ij}$ is the vector distance between node $i$ and node $j$, and $l_{ij}$ and $k_{ij}$ are, respectively the rest length and the stiffness of the spring connecting the two nodes.

The force propagation method is not ideal for global deformation due to the complexity involved. However, the force propagation method is best suited for simulations where deformation primarily occurs in localized regions [77]. Local deformation means that the deformation is limited to a relatively small region of the object (poking, nipping, pressing or cutting) [73].

Typically, linear elastic theory is used in mass-spring models, and this does not match the nonlinear properties of biological tissue. However, small deformations in biological soft tissue do reflect the deformation induced by spring models. Conversely, large deformations in biomechanical tissue are nonlinear and do not agree with the results of spring models.
6.5 Finite Element Models

Delingette and Ayache reported that the finite element method (FEM) is the most popular technique for the computation of structure deformation based on the elasticity theory [64]. Like the mass-spring models, finite element models are primarily based on linear elastic theory. Models can be static or dynamic based depending on whether inertial and viscous effects are considered. Static FEM models can’t simulate time-dependent effects such as viscoelasticity. However, only a few research groups are using dynamic models due to the high computational cost.

The use of small time steps for spring models limits their ability to model a large range of dynamic behaviors. Using the finite element method, larger time steps can be used. Consequently, the range of possible dynamic behaviors for spring models is more limited than that of finite element models [74]. Finite element models are well suited to compute accurate and complex deformation of soft tissue.

Gibson and Mirtich outlined the basic steps of the FEM [78]. First, the equilibrium equations must be derived from the potential energy equation. Second, the appropriate finite elements (in most cases, tetrahedral) are selected. Third, the equilibrium equations are re-expressed in terms of interpolation functions for each element. Fourth, the equilibrium equations for all the elements are combined into a single system. And finally, the node displacements and interpolation functions are used to calculate the displacement values for all the points within the element. In general, finite element models are computationally expensive.

Model geometry motivated the use of tetrahedra rather than hexahedra [64]. For example, meshing most anatomical structures with hexahedra is known to be especially
difficult for highly curved structures or circumvoluted parts such as the liver. In order to
obtain a smooth surface, fewer tetrahedra would be required than hexahedra.
Furthermore, several commercial and academic software programs exist for tetrahedral
topology.

A second motivation for using tetrahedral elements is related to cutting soft tissue.
Cutting actions, which are discussed in more detail later in this thesis, usually need to
remodel or re-mesh the local elements. However, hexahedral meshes required that new
element types (such as prismatic elements) be added to the model during cutting [64].
Consequently, this task would make the algorithms more complex.

The node vector $X$ represents a finite element model. For static computation, the
stress-strain relationship leads to $f(X)=0$, whereas for dynamic computation, the
following Newtonian formulation is often used [74]:

$$m \ddot{X} + \gamma \dot{X} + f(X) = 0 ,$$ (4)

The integration of the differential equation can be performed semi-implicitly or
explicitly. In general, implicit schemes are unconditionally stable, whereas explicit
schemes are conditionally stable. Therefore, smaller time steps must be used with
explicit schemes.

The implicit methods require a solution of a set of non-linear algebraic equations at
each time step. Iterations are also needed for each time step to control the error and
prevent divergence. Therefore, the number of numerical operations per time step can be
three orders of magnitude larger than for explicit integration.

On the other hand, the explicit methods are only conditionally stable. Normally a
severe restriction on the size of the time step has to be included to obtain satisfactory
simulation results [79]. Some researchers have found that the critical time step is equal to
the smallest characteristic length \( L_e \) of an element in the mesh divided by the dilatational
wave speed \( c \) [59].

In viscoelastic materials, an instantaneous elastic response occurs upon loading,
followed by a slow and continuous change in the response at a decreasing rate [80]. One
way to derive a constitutive relationship for linear viscoelastic materials is to assume that
a Boltzmann superposition of strain increments can be applied to viscoelastic materials.

Consider an arbitrary strain input obtained through superposition of small strain
increments [80]:

\[
\varepsilon(t) = \sum_{j=1}^{N} \Delta \varepsilon_j = \left[ \int_0^t d[\varepsilon(s)] \right], \tag{5}
\]

where \( s \) is any arbitrary past time between 0 and \( t \) when we apply a constant strain. Each of these strain increments is related to stress increments by Hooke’s law. The stress
increments relax according to the time dependency of the stress relaxation function \( E(t) \).
Taking the appropriate limit results in the following constitutive law [80]:

\[
\sigma(t) = \int_0^t E(t-s) \frac{\partial \varepsilon(s)}{\partial s} ds, \tag{6}
\]

The generalized Maxwell solid typically models viscoelastic materials, which is a
combination of springs and dashpots. This type of model results in a Prony series
expression for the stress relaxation function in the form:

\[
E(t) = E_\infty + \sum_{j=1}^{N} E_j e^{\left(-\frac{t}{\tau_j}\right)}, \tag{7}
\]

where \( N \) is the number of Maxwell elements, \( E_j \) is the elastic coefficient (\( E_\infty \) is the long-
term elastic modulus corresponding to the system’s steady-state elastic response), and \( \tau_j \)
is the relaxation time related to the damping coefficients and dashpots. Experimental results [80] place the long-term elastic modulus at 12.879 ± 2.95 kPa. This value corresponds to the effective linear elastic modulus of pig liver, and shows a good agreement with the values that Ottensmeyer obtained.

Sedef and Basdogan have also investigated the relaxation properties of biological tissue [80]. The force relaxation response of a pig’s liver for different loading rates lasts approximately 30 seconds. In addition, it can be assumed that the loading influences only the nodes around the contacted node within a finite radius of influence (ROI). This assumption improves the computational efficiency of the algorithm without limiting the deformation behavior.

Without any constraints the FEM’s mesh floats in space and can occupy an infinite number of positions. In order to solve a linear elastic system, there should be only one possible position. Therefore, Bro-Nielson suggested the displacement of at least three nodes must be fixed [81].

Cash et al. derived boundary conditions from knowledge of the forces applied to the liver within surgery along with information from the intra-operative data [82]. Their approach yielded three different types of boundary conditions. The first sets of boundary conditions are categorized as “fixed.” Typically, obscured regions of the right lobe that rest against other parts of the viscera belong to the fixed category. “Stress free” boundary conditions are the second category, which represent regions unrestricted by force. The final type is “closest point” boundary conditions. These nodes play the most significant role in modeling the deformation and are considered a mixed boundary condition.
In general, FEM can be simulated fully or using a condensed method, which requires less computation. The choice depends upon the requirements of the application. For example, because the condensed stiffness matrix is one step further in refinement than the standard stiffness matrix, it is more difficult to change [81]. Therefore, if a change in topology is necessary, then a change in the stiffness matrix is also required. This is usually the case when modeling cuts or incisions, which makes the extra computations of the full simulation necessary.

6.6 Free Form Deformation Models

Most deformation approaches alter the geometry of the deformed object without affecting its topology. This can become restrictive when a designer wishes to incorporate holes or tears into an existing model [83]. A deformation tool that can potentially modify the topology of a model as part of the deformation specification will offer diverse applications as a modeling tool for surgical applications.

Free form deformation (FFD) is a widely accepted general deformation tool. However, a significant shortcoming of FFD stems from the fact that it maps a box-shaped domain into a contorted-box in Euclidean space [83]. For example, when used to deform objects of complex geometry, the deformation function only remotely resembles the deformed object. Consequently, it becomes more difficult to control the final shape of the deformed object by manipulating the control points of the deformation function.

These problems with the final shape can be addressed using extended FFD (EFFD). EFFD combines multiple FFD volumes to construct a single deformation function that better resembles the shape of the deformed model [84]. MacCracken and Joy used
arbitrary topology FFD based on subdivision volumes for free-form deformation [85]. This method improves the localization of the deformation.

Discontinuous free form deformation (DFFD) will deform the model properly while automatically allowing it to split and re-form at the proper locations [83]. One DFFD application splits a deformable object then wraps it around an obstacle in the scene. A second application demonstrates DFFD as a general framework for intersecting cuts into the surfaces of geometric models.

6.7 Volume Models

Frisken-Gibson reported that volumetric representations have advantages over surface-based models when modeling object deformation because they allow the interior structure to impact the physics of object interactions [86]. However, volumetric objects consist of a large number of elements. Consequently, deformation techniques are computationally expensive.

By itself, the ChainMail algorithm, which is implemented in many volume models, has difficulties in relating the deformation of the organ surface to the internal structures [87]. It is also difficult to make incisions and assure the user of quantitative resection. Therefore, Gibson [88] investigated an alternative two-process approach to volumetric models because using mass-spring methods and FEM is too computationally expensive. The first process implemented the ChainMail algorithm, which provides an initial estimate of the new shape that is based on geometric constraints. It guarantees a plausible object shape in one time step even for relatively large deformations and large models. The second process relaxed the shape of the approximate deformation over
several time steps. The behavior of the material was determined both by the geometric constraints and by parameters in the elastic relaxation process.

Inertial behavior occurred because the elements in the ChainMail process do not move unless they violate constraints between local neighbors and the move only minimum distances in order to satisfy the violated constraints [86]. Damping occurred because the iterative elastic relaxation process was a closed negative feedback system in which element positions are adjusted in small steps toward an optimal position. For example, when the step size (gain of the feedback system) was small, the system is more damped. Likewise, an increase in step size would result in a critically under-damped system.

ChainMail uses a linked data structure [86]. For example, each element is linked to its six nearest neighbors and disturbances are propagated through the system via these links. If the displacement of an element violates the constraints on the links, then the affected neighbors are moved to the closest position where the link constraints are again satisfied. After each element is moved, its links to unmoved neighbors are checked and corrected if necessary. The disturbance is fully propagated through the volume in one time step.

The ChainMail algorithm produces a deformed shape that satisfies geometric constraints but doesn’t necessarily have an optimal energy configuration [86]. Therefore, an elastic relaxation process is applied to locally adjust relative element positions and reduce the system energy, which depends on the distances between object elements.

There are a number of limitations associated with the ChainMail deformation technique. For example, it does not model volume preservation. This becomes a
problem because living tissue is considered to be incompressible. Currently, volume preservation is addressed in the elastic relaxation step. However, researchers are working on a method to include volume preservation in the ChainMail process [86]. One suggested solution is to add diagonal links between elements.

A second limitation is it only models homogeneous materials at a single control point. Both issues have been addressed [89] by ordering lists of movement candidates at each step in the ChainMail algorithm. This ensures that elements with the largest violations on their link lengths are considered first. However, while this solution allows the consideration of heterogeneous materials and multiple contact points, the need to maintain an ordered list of movement candidates slows the algorithm’s processing speed.

One volumetric mass-spring system presented consists of a model subdivided into tetrahedra [75]. The vertices of each tetrahedron are assigned mass points, and springs and dampers connect the mass points. Newton’s law of dynamics, where the sum of internal and external forces equals the product of the mass at each point and its respective acceleration, governs the system. The internal forces are the resultant of the tensions of the springs and dampers linking $P_i$ to its neighbors $P_j$ [75]:

$$ F_{in}(P_i) = \sum_j \left[ k_{ij}(r_{ij} - |v_{ij}|) - c_{ij} \frac{v_{ij} \cdot r_{ij}}{|v_{ij}|} \right] \frac{r_{ij}}{|v_{ij}|}, \quad (8) $$

where $r_{ij}$ and $v_{ij}$ are the relative position and relative velocity of the point $P_i$ with respect to its neighbor $P_j$. The stiffness, damping and natural length of the spring between $P_i$ and $P_j$ are given by $k_{ij}$, $c_{ij}$, and $l_{ij}$, respectively.

In this approach, the model’s physical parameters, such as the point masses and the spring values, strongly depend on the mesh topology. These parameters must be updated
after every refinement or simplification during the re-meshing process. The computation depends on the number of parameters to be updated, and can quickly become expensive.

### 6.8 Non-Linear Elastic Models

Strictly speaking, all systems in reality are nonlinear systems [90]. Linear systems are only to simplify computation and to derive an idealized model. Consequently, due to the physical properties of soft tissue, the use of linear spring deformation may lead to some distortion of the result.

Non-linear elasticity allows for much more realistic deformations. However, it is more computationally expensive than linear elasticity [69]. In nonlinear approaches, strain is nonlinear in displacement and velocity, while stress is nonlinear in strain [91]. The nonlinearity of strain in the velocity field is viscosity, which appears when there is creep during deformation.

Previous experimental characterizations have revealed that viscous effects should not be ignored for an accurate description of the mechanical properties of biological tissues. Viscoelastic material properties are also nonlinear. However, in an effort to simplify the governing equations of the model, viscoelasticity can be introduced into a model by using a simple linear relation [92]. For example, Schwartz et al. introduce a viscous force that is proportional to the speed of deformation and to a viscosity coefficient. The effect of the viscous term is an increase in tissue resistance at higher compression rates [92]. The force appears as soon as compression starts if velocity is constant during the simulation.

### 6.9 Fundamental Tissue Cutting Models

Little is known about the stress/strain relationship that occurs during and after cutting. However, a basic assumption is that the physical properties of the tissue are only
modified locally [74]. For simplicity, cutting is generally restricted to the boundaries of
the model. Also, during cutting, it is necessary to update the visible faces to represent the
changes in local topography.

When a cutting blade enters a piece of tissue, the following two important interactions
occur [93]. First, the blade separates the tissue and penetrates into the interior. Second,
the forces imposed by the blade deform the tissue. The friction of the cutting blade
establishes the relations of these two interactions.

The friction occurring during a cut can be separated into two different forces [93]. 1) As
long as the force applied to the blade is smaller than a given tissue-dependent
threshold, the blade will not open the tissue. This force may be called the static cut
friction. 2) If the magnitude of the force exceeds the threshold, the blade will start
cutting. Note that the friction imposed in this case, the dynamic cut friction, is usually
smaller than the static cut friction.

Pre-loading and residual stress significantly affect the cutting of biological tissues
[94]. Such tissues commonly have a so-called J-shape stress-strain curve. For a J-shape
material, the change of stress is very smooth for a range of strain at the beginning of
deformation, but later the stress-strain curve experiences a large slope. Pre-deformation
can bring the state of deformation in a biological material to the large slope region, where
a small deflection caused by a cutting tool can generate a level of force high enough to
create fracture.

Cutting with surgical scissors involves two distinct phenomena: deformation and
fracture [95]. The scissors cause deformation when the sharp edges of the blades
compress and/or deflect the material. The fracture occurs at the front of a crack
developed in the material. It is assumed that the material is initially free from residual stress and external pre-loading. Researchers further assume that purely nonlinear deformation occurs in the material, except in a small region around the crack front where elastic-plastic fracture occurs.

Cutting in spring meshes can be implemented in two ways. In the first approach, the spring is simply removed [96-99]. In the second approach, Neumann et al. split a colliding spring [100]. A tissue laceration occurs when an excessive stress is present on some section of the soft object [101]. In a mass-spring model, the stress is directly represented by the elongation of the springs. Therefore, the obvious manner to implement lacerations is to assign to each spring a limit on elongation, and to impose that it breaks down when such limit is reached. Lacerations should start and propagate from the surface.

6.10 Cutting Finite Element Models

Implementing cutting procedures in FE systems causes two major problems [81]. The first is related to the geometric modification of the mesh. These modifications are needed to ensure a smooth cut when rendered using computer graphics. In general, it is necessary to refine the mesh around the cut, and the geometric aspects of this become difficult.

The second problem concerns the necessary change of the stiffness matrix and the linear system. For explicit methods this is simple [81]. However, implicit systems require an inverse stiffness matrix, and updating an inverted matrix is more difficult.

When stresses at a given node exceed some limit, they produce a discontinuity that results in fracture behavior. However, accurate modeling of arbitrary cutting through an
FE model would require reworking the equations governing the system [86]. The model would have to be re-meshed at each intervention as well to provide a higher resolution mesh at high stress points (knife tip).

According to fracture mechanics, an object may be broken due to two different types of failures. Tensile failure corresponds to loading normal to the failure surface. If the failure is produced by pushing rather than pulling then it is considered a compressive failure. The other type of fracture, called shear failure, corresponds to the loading tangential to the failure surface.

Nienhuys and van der Stappen’s approach to cutting FE models involved a method that does not inflate the mesh size [102]. This was done by applying cuts only to the faces. This bounds the size of the mesh, and helps keep the geometry of the mesh uniform. The idea behind this method was that mesh modifications should avoid creating more elements, which would increase the complexity of the model. Likewise, modifications should avoid creating elements with varying sizes.

There are only five topologically different cases in which a tetrahedron can be cut during an incision [103]. For each of the five cases the researchers first store the set of actions required to establish the new mesh in a lookup-table entry. These actions include the insertion of new nodes, the assignment of their connectivity, and the insertion of new faces. By rotating and mirroring these five cases they get all the possible combinations of edge intersections necessary to complement the lookup-table.

To avoid individual subdivision procedures for each of the five cases and to simplify implementation, researchers propose a generalized subdivision scheme that divides a tetrahedron into a fixed number of 17 smaller units that are independent of the topology
of the cut [103]. For all edges and faces, the correct geometry of a cut surface is computed by replacing the indicated edge and face midpoints by the current intersection points. Finally, by referencing the midpoints twice and the face midpoints three times, the subdivision scheme can be implemented as a pre-split tetrahedron.

There are a couple of limitations to the generic subdivision procedure [103]. The first one relates to the potential existence of hanging nodes that have no connection to adjacent tetrahedra. This may lead to cracks in the visual representation. It is possible to solve for all hanging nodes by splitting adjacent tetrahedral appropriately. However, researchers must balance accuracy against computational costs in order to determine whether the splitting operation is necessary.

The second limitation is related to the order of the procedures [103]. For example, the splitting procedure is invoked after completing the cut. This might lead to minor positional and visual discontinuities.

This scheme works well in practice. However, it may lead to a rapidly increasing number of tetrahedra for large cuts [93] because a large number of elements would be subdivided. In addition, cracks may arise in the physical representation of the tissue. A revision to their method confined the newly created pieces to the actual tetrahedron by introducing cut-specific subdivision patterns. For example, the procedure only subdivided the edges and faces that were part of the cut surface.

The localization of the subdivision was clearly visible [93]. All faces and edges that were not intersected by the cut surface remain unchanged. An additional node was inserted into every completely cut face, but it was not necessarily needed. However, researchers found that this operation provided two advantages. First, it led to more
symmetry in the subdivision patterns. Second, it created additional degrees of freedom for the cut surface improving the results of the subsequent numerical relaxation.

Elastic tissue behavior forced the cut surface to open during this relaxation process [93]. In order to provide the necessary degrees of freedom, all newly inserted mass nodes, edges, and faces of the cut surface that do not belong to the boundary were inserted twice. These new subdivision rules have been extended into the lookup-table previously discussed [103].

Besides mesh subdivision, an alternative cutting simulation technique is to modify the shape of mesh elements by dislocating the nodes, so as to adapt to the trajectory of the cutting tool and generate smooth cut surfaces [104]. This technique, called node snapping, attempts to avoid the increase in mesh complexity due to subdivision. However, some problems arise with the use of this method. First, node snapping may cause undesirable changes to the external shape of the mesh. Also, an extra processing step is also needed to eliminate degenerate elements.

Many methods of modifying objects do not split elements until the cut has been completed [105]. When the average element size is quite large, this introduces a noticeable lag into the cutting process. However, progressive cutting generates a minimal amount of subdivision of a partially cut tetrahedron in an effort to reduce this lag.

The general procedure for progressive cutting utilizes a temporary subdivision of each partially cut element [105]. First, any temporary face intersections caused by cutting edge are updated for each partially cut tetrahedron. A temporary face intersection occurs when the cutting edge intersects a face. Then, the modified topology of the partially cut element is checked for any changes. A change occurs when the element is
first cut into, or when the cutting edge or tip passes through another edge or face. If the topology has changed, new minimal sets of temporary tetrahedra are created and all of the old temporary tetrahedra are removed. If the modified topology has not been changed, then the temporary elements are updated using the new positions of any temporary face intersections. Finally, once the cutting edge leaves an element and the cut is completed, the temporary elements are removed, and a final subdivision is created.

There are two important issues unique to progressive cutting that must be addressed [105]: how to deal with the cutting edge being within the interior of an element; and how to deal with two intersections on one face. When the cutting edge is within the interior of an element, ideally the model should be able to open up along the cut of the blade. However, given the nature of the subdivision for a generic cut, this would not be possible. Likewise, if no intermediate nodes are inserted between the two face intersections, a straight line will always connect them. Also, when there are two intersections on one face, none of the triangles generated on that original face overlap. If this occurs, then the modified topology of the partially cut tetrahedron has changed, and a new set of tetrahedra will be created.

Alternatively, the DFFD algorithm can also be treated as a local mesh-splitting algorithm that cuts the surface of meshes [83]. The algorithm also models the shape and deepness of the cut produced by the shape of the cutting tool. Moreover, by properly designing the DFFD function, the algorithm can simulate the time-response of the deformable object in the vicinity of the cut.

Applying the DFFD to local areas of a model has two direct benefits [83]. First, since only part of the model undergoes a deformation, the algorithm can achieve faster
computation rates. And second, the local version of the algorithm lets the designer specify arbitrarily shaped incisions.

Like surface models, volumetric models also have the tendency to produce self-intersections during cutting. However, linked volumes are well suited to the modification of object topology [86]. They can be cut, carved, and torn by removing elements or by breaking the links between elements. These modifications can be made at higher resolutions because linked volume models have more elements than FEM or mass-spring models.

During cutting, the tool can remove links, as well as elements that are encountered along its path as it passes through the object. Intersections between the cutting path and element links are detected by moving the cutting tool volume through the occupancy map and checking for collisions. If the tool path encounters an element, the element is removed and appropriate neighbor connections are removed. If the tool passes through a link between two elements, the connection between those two elements is removed in both elements.

An alternative approach uses a splitting technique for three-dimensional volumetric mesh [103]. Biesler et al. subdivided any tetrahedron intersected by the cutting tool. These cuts added lots of small volumetric elements, and consequently, the mesh size tended to explode. This disadvantage caused a deterioration of the numerical stability of the spring relaxation scheme. In addition, rounding errors could cause dangling nodes.

Many of the re-meshing approaches proposed for volumetric deformable models rely on a multi-resolution presentation, which is built offline in a preprocessing step [75]. However, these methods have drawbacks. For example, large memory is required to
store the multi-resolution data representation. Also, the data structure strictly depends on the object’s topology, which is supposed to be invariable, and thus prohibits structural modifications.

### 6.11 Hybrid Models

The key idea behind hybrid models is to reduce computation requirements by modeling the local and global areas differently. However, hybrid models present some drawbacks as well. For example, ensuring continuity and smoothness on the interconnecting boundary between two sections is often not straightforward [101]. Moreover, subsections affected by the dynamic modification actions must be precisely planned in advance, and the boundary of a subsection must be precisely defined as well.

This approach fundamentally breaks into two different types of anatomical models [106]. The first model is of the anatomical structures where the surgery occurs (pre-computed). On these structures, tearing and cutting need to be simulated. This is especially feasible when the surgical operation to be simulated is well known when constructing the model. The second type of model is of the anatomical structures that only need to visualized or deformed in order to contribute to the visual realism of the simulation. The approach can also be applied to the same structure.

Some researchers have developed a node snapping, mesh-less hybrid model in order to achieve realistic simulation of surgical cutting. Physically based computations were performed using the mesh-less numerical technique up to the point of tissue rupture [107]. Thereafter, a node-snapping technique was used for edge following and simulation of progressive cutting at real-time rates [108].
The point-associated finite field (PAFF) modeling technique is a mesh-less computational scheme in which partial differential equations may be solved on geometrically complex domains discretized using a scattered distribution of points. Unlike mass-spring models, which require hundreds of thousands of parameters to be empirically determined, the advantage of PAFF is that only a few well-defined experiments are needed to obtain the material parameters [109]. And unlike FEM, large tissue deformations are particularly easy to handle using PAFF because interpolation functions are compactly supported on spherical subdomains, which may intersect and overlap and are not constrained to each other as in FEM.

The cut is opened by first splitting each node that lies on the cut into two nodes which are displaced perpendicular to the cut by equal amounts [109]. However, a major problem is that the resolution of the cut depends on the amount of refinement in the underlying polygon model. In order to overcome this problem and improve visual realism, researchers implemented a local subdivision algorithm. After the first contact of the cutting tool with the organ model, the surface triangles within a certain “radius of influence” are subdivided with different levels of detail.

### 6.12 Puncture and Cutting Forces

Little information can be found in the literature on the optimum forces and deformations required for safe and efficient surgical dissection. Carter et al. measured both the puncture force and the displacement of the puncture site that takes place before puncture occurs [110]. It was recognized that the resistance to puncturing the liver derives mainly from the tough superficial region.
The membrane of porcine liver contains a large number of collagen fibers (elastic modulus = 1000 MPa, rupture at 10-15% elongation) and virtually no elastin fibers, causing it to be much less extensible. It was found that increasing pre-strain caused a decrease in the displacement before puncture, but it had no effect on the required puncture force [110]. When the blade is pressed against the capsule, a contact reaction force is generated, which is equal and opposite to the force applied by the blade. The reaction force is a component of the tissue tension resolved in the direction of the blade movement, and it increases as the displacement increases until puncture occurs. Adding pre-strain to the tissue increases the reaction force, and therefore, the puncture force will be reached at a smaller displacement.

The cutting force is that which is necessary for the tool tip to slice through the tissue. Okamura et al. [111] claimed that this force was a combination of cutting force and tissue stiffness at the tip of the cutting tool, since the tip compresses the tissue before fracture. Subtracting the previously determined friction force from the total force isolated the cutting force. It is difficult to accurately estimate the cutting force due to the possibility of collisions with internal vessels.
Chapter 7

7 Tissue Simulation

7.1 Adding Tissue to the Biopsy Grasper Simulation

The optimal mechanical design of the biopsy grasper was discussed in detail in Chapter 5. Design characteristics, such as Nitinol ribbon thickness and profile angle, were modeled and analyzed for the mechanical operation of the biopsy grasper. The simulation discussed in Chapter 5 did not include any interaction between the biopsy grasper and a liver tissue model.

As stated in Chapter 6, modeling biological tissue, such as the liver, is very complicated. For the purpose of this thesis, a rough model of the liver tissue is used to complete the analysis of biopsy grasper design parameters. The creation of a realistic and detailed tissue model is beyond the scope of this thesis. However, a simple model that behaves in accordance with experimental observations can be used to determine the effects of different design parameters of the biopsy grasper.
In order to model the large deformations that are present during a liver biopsy, a small mesh size must be used in the model. Adding all of these nodes and elements complicate the simulation. Therefore, all unnecessary components of the biopsy grasper are left out of the tissue simulations. Specifically, only the top and bottom blades are moved using displacement boundary conditions. Moreover, the blades are modeled as rigid elements. Because the focus of this study is the interaction between biopsy jaws and the tissue, this simplified model is sufficient for determining optimal design and placement of the jaws for the most efficient tissue sampling.

7.2 The Elastic Tissue Model

The review of literature on soft tissue modeling in Chapter 6 suggested that there is a wide range of values accepted as material properties for liver tissue. For this model material properties are derived from a study by Ottensmeyer et al. in 2001. The Young’s modulus is 15 kPa, and the yield stress is 2.5 MPa. For the simulations discussed in this chapter, the tissue is modeled as an elastic material. While it is widely known, that biological soft tissue is not purely elastic, these preliminary tests serve as a starting point and control for the analysis of the interaction between the biopsy grasper and liver tissue.

A rectangular block 10 mm x 10 mm x 1 mm thick, which is relatively large compared to the size of the biopsy grasper jaws, was used to represent the liver tissue. The mesh, shown in Figure 7-1, is constructed out of quadratic, hybrid brick elements that had 20 nodes each. Moreover, only a small area of the tissue deforms during the simulation due to the low stiffness of the material properties. Therefore, the fine mesh (outlined in yellow with a mesh size of 0.1 mm) is not needed throughout the entire liver model. Boundary conditions (highlighted in red) are placed along the edges and areas of the base
of the liver tissue. While the area of biopsy sampling is relatively free to move, the majority of the liver is undisturbed as the biopsy grasper closes.

![Mesh of Liver Model](image)

**Figure 7-1: Mesh of Liver Model**

Once again the interaction properties are modeled as “hard contact” in the normal sense. However, due to the large deformations that are present in very small areas, a node to surface interaction is used in this simulation. Moreover, only a few elements are in contact with the thin blades of the biopsy grasper. However, each element consists of 20 nodes, and as a result, a node to surface interaction property is preferable so that more points of interaction are modeled during the simulation.
For the same reasons, some small sliding is allowed between the master (top and bottom blades) and slave (liver tissue) surfaces. The small sliding allows the surfaces to overlap slightly during the simulation, which reduces the risk of divergence during simulation. This model is an adequate representation of observations made during lab experiments.

7.3 Positioning the Biopsy Blades on the Liver Tissue

During the \textit{in vivo} navigation and use of the biopsy grasper, it will be extremely difficult to accurately and repeatedly place the biopsy grasper in the perfect position over the edge of the liver tissue. For example, organs inside the abdominal cavity experience small movements due to the respiratory motion during surgery. However, in order to extensively evaluate the biopsy jaws, a common starting place must be used so that simulation results can be compared.

The top blade and bottom blade are concentric in order to ensure a uniform interaction between the liver tissue and blades. By ensuring that the clearance between the blades is uniform along the entire top jaw, any unexpected stress concentrations are avoided. A schematic of the position of the jaws is shown in Figure 7-2. Initially, the jaws are positioned so that the back edge of the top blade is on the edge of the liver tissue. Positions in which the top blade is not entirely in contact with the liver tissue are not investigated. This is due to the fact the biopsy grasper performs its best when the overlapping area between the top and bottom blades is maximized.
Simulations are performed while moving the position of the blades inward. The maximum stress during the simulation is shown in Table 7-1. The simulations result in a maximum stress when the top blade was 0.2 mm inward from the edge of the liver tissue, and this position is considered the best position for the simulation. Therefore, this position of the back edge of the top jaw (0.2 mm inward from the edge of the liver tissue) is used for all the other simulations discussed in this chapter.

![Schematic of Jaw Position on Liver Tissue](image)

**Table 7-1:** Simulation Results with Varying Jaw Position

<table>
<thead>
<tr>
<th>Jaw Position from Edge</th>
<th>Maximum Stress</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.00 mm</td>
<td>0.1953 MPa</td>
</tr>
<tr>
<td>0.1 mm inward</td>
<td>0.1964 MPa</td>
</tr>
<tr>
<td><strong>0.2 mm inward</strong></td>
<td><strong>0.1976 MPa</strong></td>
</tr>
<tr>
<td>0.3 mm inward</td>
<td>0.1973 MPa</td>
</tr>
</tbody>
</table>
7.4 The Hyper-Elastic Tissue Model

The elastic tissue model does not accurately represent the mechanical response of living tissue. However, the elastic model is effective in its intent to evaluate the positioning of the biopsy blades on the liver tissue. The fracture mechanics of biological soft tissue are very complex, and the elastic model allows the tissue to undergo large strains during the simulation. This section discusses a hyper-elastic tissue model, which is a more realistic representation of living tissue.

The hyper-elastic tissue model is incompressible. As strain increases, the stiffness also increases. This model more closely represents the stiffness that is expected with living liver tissue. For the purpose of these simulations, a previously published hyper-elastic tissue model [112] is used to evaluate the biopsy grasper. Data from Zhong’s experimental measurements are implemented into the Abaqus software. Simulated material evaluations are performed using Abaqus to find a stable and approximate material model. A two-parameter reduced polynomial hyper-elastic material model is shown in Figure 7-3. The nominal stress units are in MPa. The modulus used in the elastic model discussed in Section 7-2 matches up with a strain of approximately 0.35 in Figure 7-3. Consequently, the stiffness of the hyper-elastic model is greater under large strains than the stiffness of the previous elastic model. The Abaqus parameters for the fitted reduced polynomial curve are: C10 = 0.0023354 and C20 = 0.0106185.
The distribution of the mesh in the hyper-elastic model is slightly different. Because the hyper-elastic model is incompressible, a larger portion of the tissue model experiences deformations. Consequently, a larger area of fine mesh is needed for the hyper-elastic tissue model. Also, the amount of fixed boundary conditions on the liver model is reduced.

For the hyper-elastic mesh, the elements are hexagonal, linear, hybrid brick elements with reduced integration. Each element consists of 8 nodes. Furthermore, the increased number of elements undergoing deformation requires a reduction in the boundary conditions. For the hyper-elastic model, fixed boundary conditions are applied to only the back edge of the tissue model and to a portion of the bottom face of the model. Other various boundary conditions lead to convergence problems due to large element displacements necessary to solve the hyper-elastic response of the tissue model. Finally,
the hyper-elastic tissue model also utilizes the same interaction properties as the elastic tissue model (discussed in Section 7.2).

7.5 Matched Jaws vs. Overlapping Jaws

Traditional biopsy tools use jaws that overlap in order to collect large samples and cut more tissue during the sampling process. On the other hand, laparoscopic biopsy tools are typically constructed with jaws that do not overlap. One of the major redesign characteristics of the biopsy grasper presented in this thesis is the implementation of overlapping jaws.

Consequently, the first investigation to perform using the tissue model is to investigate the difference between jaws that overlap and jaws that do not. An analysis of the matched jaws is shown in Figure 7-4, and a maximum stress of 0.05172 MPa is calculated during the simulation. The figures illustrating deformation in this chapter indicate severe distortions in the model elements. Refining the model by using a smaller mesh size or an adaptive mesh is beyond the scope of this thesis. Furthermore, these figures are presented solely as an aid for visualizing the effects of different grasper design parameters.

Figure 7-4: Elastic Model Using Matched Biopsy Jaws
On the other hand, Figure 7-5 shows the simulation results when the jaws overlap, as designed for this study. The maximum stress calculated increases to 0.2061 MPa, which is approximately a factor of four larger than the maximum stress for matched jaws. Also, the distortion and deformation of the tissue model is much more severe near where the blades overlap. The combination of greater stress, displacement, and distortion indicates that more damage is done to the tissue model. Moreover, areas of fracture and tearing are more likely to occur with overlapping jaws.

![Elastic Model Using Overlapped Jaws](image)

**Figure 7-5:** Elastic Model Using Overlapped Jaws

A similar comparison is made using the hyper-elastic model. The results are similar to the elastic model simulations, but the maximum stress calculations in the hyper-elastic model are much higher than the elastic model. This is a result of the increased stiffness of the hyper-elastic model as deformation increases.

Because the incompressible hyper-elastic tissue model diverges rapidly under larger deformations, the displacement boundary conditions are reduced. For example, when the blades are matched up, the bottom and top blades are only displaced 0.27 mm and 0.92
mm respectively. This is approximately 54% of the displacement boundary condition applied to the blades in the elastic tissue model. The maximum stress calculated with the matched jaws is 0.07264 MPa. However, the maximum stress calculated with overlapping jaws increases to 0.2349 MPa. The difference between matched and overlapping jaws is similar to the difference calculated using the elastic tissue model. Therefore, the hyper-elastic and elastic models both suggest that an overlapping biopsy grasper is able to apply more stress to the tissue model.

7.6 Clearance Between Top and Bottom Jaws

A tight clearance between the top and bottom jaws is desirable to create a large shear force and initiate the tearing of tissue. The optimal clearance between those jaws is discussed in this section. Three different methods were used to vary the jaw clearance during simulations. First, the diameter of the bottom jaw is decreased in order to produce more clearance. Second, the diameter of the top jaw is increased to produce a similar effect. Finally, the position of the top jaw with respect to the bottom jaw is also adjusted to create more clearance between the blades.

The first set of analyses on the biopsy blade design is to investigate the effect that the diameter of the bottom blade has on the ability to cut tissue. The prototype design has a top blade outer diameter of 4.22 mm and an inner diameter of 3.46 mm. The bottom jaw has an outer diameter of 3.40 mm and an inner diameter of 2.70 mm. During these simulations, the outer diameter is reduced to 2.80 mm.

The simulation of the 3.40 mm bottom blade is shown in Figure 7-6. For comparison, the results for a bottom blade diameter reduced to 2.80 mm are shown in Figure 7-7. When the outer diameter of the bottom blade is decreased, the clearance between the top
and bottom blades increases. Differences in the liver tissue model are seen when comparing Figure 7-6 and 7-7. When there is more clearance between the top and bottom blades, the tissue model appears to bend and deform between the two blades. There is clearly more deformation and distortion due to shearing effects visible in Figure 7-6.

**Figure 7-6:** Elastic Model with 3.40 mm Bottom Blade Diameter

**Figure 7-7:** Elastic Model with 2.80 mm Bottom Blade Diameter
Due to the incompressibility of the hyper-elastic model, the displacement boundary conditions are decreased from those used in the elastic model. For these simulations, the bottom blade and top blades are displaced 0.31 mm and 1.08 mm respectively. These displacement boundary conditions are approximately 64% of the displacement implemented in the elastic model simulations.

The next investigation varies the diameter of the top blade while the bottom blade diameter is held constant. The diameter of the bottom blade is returned to 3.40 mm for this investigation. Meanwhile, the outer diameter of the top blade is increased from 4.22 mm to 4.80 mm.

The 4.22 mm diameter top blade produces the most stress on the elastic tissue model, and the simulation results are shown in Figure 7-8. The distortion and displacement of the tissue is much greater than the simulation results when a larger diameter top blade is used. As the clearance between the top and bottom blades increases, the maximum stress calculated by the simulation decreases. By comparison, the simulation results of the 4.60 mm diameter top blade are shown in Figure 7-9. Once again, the increase in clearance allows the tissue to deform between the biopsy blades. Moreover, it is clearly visible that the deformation and distortion of the liver tissue model is much higher with less clearance (Figure 7-8) than it is when the clearance is increased (Figure 7-9).
As the clearance between the two blades increases, the stress applied to the tissue decreases for both the elastic and hyper-elastic models. Simulation results, shown in Figure 7-10, suggest that the smallest possible clearance while maintaining overlapping
jaws produces the most stress on the tissue model. The elastic simulations calculate maximum stress values on the tissue model between roughly 0.17 and 0.19 MPa for variations in both the top and bottom blades. As the bottom blade diameter is varied, the hyper-elastic model results are similar, such that the applied stress declines at a somewhat steady rate. However, in the hyper-elastic model, the maximum stress declines much more rapidly, and it also decreases to a much smaller value than the elastic model.

![Figure 7-10: Results Comparison of Hyper-Elastic and Elastic Models](image)

The most noticeable differences between the hyper-elastic and elastic tissue models are between the clearance of 0.03 mm and 0.11 mm. For example, the elastic simulations indicate similar results as clearance increases regardless of whether the bottom blade diameter decreases or the top blade diameter increases. However, the hyper-elastic model results suggest that increasing the top blade diameter reduces the effectiveness of the biopsy grasper much more significantly than decreasing the bottom blade diameter.
This is most likely a result of the top blade displacing approximately three times greater than the bottom blade. Consequently, the stiffness of the hyper-elastic model increases dramatically. And therefore, as the clearance increases, more tissue fits between the blades, and as a result, the tissue near the grasper experiences less displacement during the simulations.

After varying the diameter of the top and bottom blades independently, the final analysis regarding clearance varies the position of the top and bottom blades with respect to one another. For this investigation the top blade diameter is 4.22 mm. However, the bottom blade diameter is reduced to 3.32 mm to avoid interference between the two blades during actuation. To illustrate, the top blade is larger than a semicircle. Consequently, if the top blade is moved while the bottom blade is held stationary, the back edges of the top blade would come into contact with the bottom blade. Furthermore, during this set of analysis, the bottom and top blades are not concentric. With the reduction of the bottom blade to a diameter of 3.32 mm, the initial clearance of these simulations is 0.07 mm, and the top blade is moved inward onto the tissue model in increments of 0.05 mm.

The tissue model for the initial clearance of 0.07 mm is shown in Figure 7-11. For comparison, the tissue model for the clearance of 0.27 mm is shown in Figure 7-12. The differences between these two figures are consistent with the simulations that varied the top and bottom blade diameters. However, one noticeable difference is visible in Figure 7-12. Because the top and bottom blades are no longer concentric, the top edge of liver tissue model remains relatively undamaged during the simulation. Therefore, it is essential that the top and bottom blades of the biopsy grasper remain concentric. This
orientation ensured a uniform cutting surface along both the top and bottom edges of the liver tissue.

**Figure 7-11:** Elastic Model with 0.07 mm Clearance

**Figure 7-12:** Elastic Model with 0.27 mm Clearance
For the hyper-elastic simulations the displacements of the bottom and top blades are 0.34 mm and 1.16 mm respectively, which is approximately 68% of the displacement used in the elastic simulations. Again, this was done to avoid divergence problems in the incompressible model. Both the elastic and hyper-elastic model results, shown in Figure 7-13, indicate that as the clearance between the blades increases, the stress applied to the tissue model decreases. However, for the hyper-elastic results, the effect of increasing the clearance between the two blades is much more significant than the results of the elastic model.

![Figure 7-13: Model Comparison When Varying Top Blade Position with Respect to Bottom Blade](image)

### 7.7 Shape of Top Blade

The semicircle design of the top blade is intended to maximize the opening of the jaws prior to actuation. If a full circle is used, the jaw opening is reduced. Consequently, it is extremely difficult to accurately and repeatedly maneuver the biopsy grasper over the edge of the liver.
Simulations are used to analyze the effect of the top blade profile. For example, the amount of wrap around by the semicircle is varied to investigate whether or not it improves the performance of the biopsy jaws. The original design has a top blade design of 62% of a circle. The amount of the circle is reduced to approximately 50% and the results are listed in Table 7-5.

<table>
<thead>
<tr>
<th>Amount of Circle</th>
<th>Maximum Stress</th>
</tr>
</thead>
<tbody>
<tr>
<td>62%</td>
<td>0.2001 MPa</td>
</tr>
<tr>
<td>61%</td>
<td>0.1965 MPa</td>
</tr>
<tr>
<td>60%</td>
<td>0.1993 MPa</td>
</tr>
<tr>
<td>57%</td>
<td>0.1953 MPa</td>
</tr>
<tr>
<td>55%</td>
<td>0.2017 MPa</td>
</tr>
<tr>
<td>53%</td>
<td>0.2144 MPa</td>
</tr>
<tr>
<td>50%</td>
<td>0.2092 MPa</td>
</tr>
</tbody>
</table>

**Table 7-5:** Simulation Results When Varying Top Blade Shape

The results show only a small variance in the stress applied to the tissue model. Therefore, it appears that the top blade shape does not have a great effect on the biopsy grasper design. However, it is important that the top blade consist of over 50% of a circle. This design allows the back edges of the top blade to wrap around the center axis of the jaws and create more area of deformation and possible tearing. The sharp back edges of the top blade serve as the primary locations for the tissue sample to begin tearing, and these points are not present if the top blade is also a full circle like the bottom blade.

### 7.8 Angle of Top Blade

One final area of consideration regarding the design of the biopsy grasper is the angle of the top blade with respect to the horizontal. Simulations are used to analyze the response of the elastic tissue model as the top blade angle of orientation is varied in the manner shown in Figure 7-14.
The top blade angle does have an important effect on the biopsy grasper design. For example, if the top blade does not match up well with the bottom blade, only small areas of the tissue would begin to fracture. Likewise, if the entire top blade does not fit tightly against the bottom jaw when the biopsy grasper is closed, the tissue sample will only be slightly torn from the liver and not completely severed. Therefore, it is important that the top blade have an orientation that allows it to match up for a tight fit with the bottom blade and jaw when the grasper is closed.

When the top blade is angled in the positive direction (Figure 7-15), the back edges of the top blade made the initial contact with the tissue model. After the biopsy grasper is closed and the top blade comes into contact with the bottom jaw (Figure 7-16), there is only a small amount of distortion and displacement present on the edge of the tissue model. It is also visibly clear that the front of the top blade would not come into contact with the bottom jaw. Consequently, it is more likely that the tissue sample would not be completely severed from the liver if the top blade is angled in a positive direction with respect to the liver.
However, when the top blade is rotated in a negative orientation, there is a significantly greater amount of distortion and displacement on the tissue model. Again, there is not a significant change in the maximum stress measured. But there is a
noticeable advantage when the front edge of the top blade makes the initial contact with the tissue model (Figure 7-17). The ribbon flexes under the force of the collar actuation to maintain a uniform contact between the top blade and bottom jaw (Figure 7-18).

**Figure 7-17:** Early Deformation of Tissue Model with Negative 10° Orientation

**Figure 7-18:** Final Deformation of Tissue Model with Negative 10° Orientation
The blade displacement boundary conditions are decreased to 60% in hyper-elastic model because the smaller contact areas resulting from the top blade angle lead to more divergence problems. Both the elastic and hyper-elastic simulation results are shown in Figure 7-19. Both results indicate that a top blade that matched perfectly with the bottom blade produces the most stress on the tissue model. However, unlike the elastic model, which remains relatively unchanged as the top blade angle varies, the hyper-elastic model results of the positive angle orientation indicate a more significant decrease in the stress applied to the tissue model. The magnitude of this difference is shown in Figure 7-19.

![Image](Figure 7-19: Model Comparison When Varying Top Blade Angle)

The effects of biopsy grasper position, blade position, blade clearance, and top blade angle are all investigated using simulations with an elastic tissue model. The analysis indicates a jaw positioning near the edge of the liver tissue is the most feasible location
for tissue collection. Data also indicate that the use of overlapping jaws result in more stress, distortion, and displacement to the tissue model. Consequently, overlapping jaws are more suitable for tissue collection than conventional laparoscopic biopsy tools. A tight clearance between the top and bottom blades is desirable. For example, a clearance between 0.05 mm and 0.20 mm results in the greatest stress values applied to the tissue model. The hyper-elastic model is incompressible and does not deform like the elastic model. Changes occur more rapidly in the hyper-elastic model than they do in the elastic model. A frame by frame representation of the simulation is shown in Figure 7-20.

Finally, the simulations indicate that the biopsy grasper is most effective when the top blade comes into flush contact with the bottom jaw when closed. While the top blade angle does not seem to affect the elastic model, the hyper-elastic simulations indicate a sharp decline in biopsy grasper performance when the back edges of the top blade initiate the contact with the tissue model.

Images from the hyper-elastic simulation are shown in Figure 7-21. The very front tip of the top blade does not show any stress applied to the tissue model because the edge of the blade does not come into contact with any nodes of the tissue model. Moreover, the sensitivity of the mesh size is especially significant in the hyper-elastic simulations. Again, Figure 8-6 indicates severe distortions in the model elements. Refining the models by using a smaller mesh size or an adaptive mesh is beyond the scope of this thesis. Furthermore, these figures are presented only as an aid for visualizing the effects of different grasper design parameters.
Figure 7-20: Elastic Tissue Model Simulation
Figure 7-21: Hyper-Elastic Tissue Simulation
Chapter 8

8 In Vivo Test Results

8.1 Sterilization

Sterility is an important design goal. It is not easy to design reusable robot components that can withstand multiple sterilization cycles [52]. A practical solution is to create disposable robotic devices that only need to be sterilized once. The modular robotic platform is intended to be a disposable device, and the low cost PolyJet manufacturing process makes this goal much more feasible. It is possible, however, for some of the end effectors or tools to be reused. Again, the modular design ensures that tools such as the biopsy grasper can easily be removed and sterilized for additional use if necessary.

Sterility tests of all of the components of the aluminum prototypes of the mobile platform were conducted using the STERRAD Sterilization System. This technique uses a combination of hydrogen peroxide vapor and low-temperature gas plasma to rapidly sterilize medical instruments without leaving toxic residues. The STERRAD system
provides a safe and effective sterilization process without the limitations or risks associated with peracetic acid, steam, and ethylene oxide gas (EtO) systems [113]. The robot motors, electronics, battery, and imager were all fully functional after sterilization. The aluminum external components (e.g., wheels and body parts) were also sterilized and subsequently tested for microbial contamination by pathologists at the University of Nebraska Medical Center. All of these tests came back negative for pathogen growth.

Additional tests were conducted using sterilized robots in three separate survivable animal cholecystectomy procedures. The robots were used to provide additional visualization capabilities during the laparoscopic procedures. The porcine models were monitored for two weeks following the procedures and then sacrificed. The results of all post-mortem visual and pathological tests showed no complications or infections [114].

8.2 Mobility

The ability of wireless robotic platforms to maneuver within the abdominal environment was demonstrated \textit{in vivo} in a porcine model. Two robots, one equipped with a sensory payload and the other with a biopsy grasper payload, were inserted into and successfully navigated the abdominal cavity. They were able to traverse all of the abdominal organs (e.g., liver, spleen, and bowel) without causing any visible tissue damage. Video recorded through a laparoscope was used to reconstruct the paths traversed by the robots, a portion of which is illustrated in Figure 8-1 for the sensor robot.
Images from the video taken with the laparoscope during *in vivo* testing are shown in Figure 8-2. The sequence illustrates the mobile modular robot platform navigating the abdominal cavity. During this particular test, the modular platform was equipped with a biopsy grasper, and the robot was moving towards the edge of the liver tissue in order to obtain a tissue sample.
Because communication to and from each individual robot used different frequencies, multiple robots could be controlled independently at the same time. At various times throughout the surgical experiments, multiple mobile robots were simultaneously controlled *in vivo* as shown in Figure 8-3, which demonstrated the ability to command and control a set of robots that could be used cooperatively. After exploring the abdominal cavity, the sensor robot was parked while it continued to monitor temperature, pressure, and relative humidity during the completion of biopsy robot and other tests.

![Figure 8-3: In Vivo Navigation of Multiple Robots (© IEEE 2009)](image)

### 8.3 Wireless Communication

The telemetry and sensor platforms have also been used in a series of *in vivo* tests in porcine models to evaluate overall system performance. First, the reliability of the telemetry system was independently examined. A master control circuit board (without the driver IC) and battery were mounted in a protective silicone tube and placed into an insufflated porcine abdominal cavity. A similar *ex vivo* transceiver board located approximately 5 m from the operating table was used to send commands to the *in vivo* transceiver. The *in vivo* device was programmed to relay the received commands to a
second *ex vivo* receiver. The data arriving at the *ex vivo* receiver were recorded to monitor the reliability of transmitted commands that made it to and from the abdominal cavity.

Successful transmission was defined as a packet completing an entire loop of communication to and from the *in vivo* transceiver. Although the communication code had not yet been optimized, approximately 90% of all packets were successfully received *in vivo* for the duration (50 minutes) of this test, shown in Figure 8-4. Reliability for *in vivo* to *ex vivo* transmission was approximately 85% during this same period. This test was subsequently repeated at a later date for a duration of 105 minutes and obtained similar results (86% and 81% reliability for *in vivo* and *ex vivo* transmissions, respectively).

![In Vivo Reception](image)

**Figure 8-4: In Vivo Wireless Transmission Reliability**
The main control board is equipped with LEDs so that operators have visual signs of communication and power during *in vivo* testing. For example, a sequence of flashing lights indicated the status of wireless communication. Likewise, an LED was used to indicate that the main board was on and functioning properly. Video snapshots from the *in vivo* wireless communication tests, shown in Figure 8-5, illustrate an LED sequence similar to the one used in the main control board of the modular robot platform.

![Video Snapshots from Wireless Communication Tests](image)

**Figure 8-5:** Video Snapshots from Wireless Communication Tests

### 8.4 Physiological Sensor Package

In a separate test, a complete physiological sensor package was integrated into the modular mobile robotic platform and the ability to monitor *in vivo* physiological parameters was evaluated. Figure 8-6 shows a typical plot of temperature (T), pressure (P), and relative humidity (RH) variations within the abdominal cavity. This telemetry was monitored and recorded at a workstation located approximately 5 m from the operating table.

The data clearly tracked significant events during the test. The temperature, initially indicating room temperature, measured a rapid rise upon insertion into the abdominal cavity. The pressure and relative humidity data also increased corresponding to the conditions within the insufflated cavity. The insufflation pressure was cycled several times during the course of the test, and those fluctuations were also apparent in the
recorded data stream. After approximately 88 minutes, the robot was removed from the abdominal cavity, and the temperature and pressure returned to *ex vivo* conditions. The relative humidity, however, continued to increase. However, a small amount of fluid was later found trapped within the body of the robot, which explained the high *ex vivo* humidity readings.

![Sensory Data from In Vivo Testing](image.png)

**Figure 8-6:** Sensory Data from *In Vivo* Testing (© IEEE 2009)

In many of the preliminary *in vivo* tests, the robot was not completely sealed prior to insertion. This was done in an effort to speed up the development phase of these modular robotic platforms. During many surgical tests on porcine models, the prototypes were inserted and removed to make adjustments or exchange parts or payloads. However, during some of the final *in vivo* tests, when prototypes were becoming finalized, the robot housing was sealed using an ultraviolet-cure epoxy common to many disposable surgical devices.
8.5 Biopsy

The biopsy robot was also used to sample hepatic tissue during the porcine surgery. The mobile robot was driven up to the point of interest along the edge of the liver, and the biopsy grasper was used to clamp onto the tissue, as shown in Figure 8-7, while being entirely remotely controlled. Small wheel movements were used to maneuver the edge of the liver into the biopsy grasper. Next, the actuation mechanism was commanded to close the jaws on the sample, and then the robot was driven away from the sample site separating the sample from the liver. Finally, the jaws were opened and the specimen was retrieved using a laparoscopic tool, as shown in Figure 8-8.

Figure 8-7: The In Vivo Biopsy Robot Sampling Hepatic Tissue (© IEEE 2009)

Figure 8-8: Hepatic Tissue Sample Removed Using the Biopsy Robot (© IEEE 2009)
Several additional specimen samples were collected to demonstrate repeatability of the device and biopsy mechanism. These samples were collected from the jaws after the robot was removed from the abdominal cavity. Once the jaws were closed and the sample removed, the mobile robot platform navigated to an area where it could be removed from the abdominal cavity. In Figures 8-9 and 8-10, the hepatic tissue sample was removed on a nearby surgical table. The mobility of the biopsy robot allows multiple tissue samples to be taken from various locations without adding any additional incisions or trocars.

Figure 8-9: Biopsy Grasper with Tissue Sample

Figure 8-10: Hepatic Tissue Sample Removed from Biopsy Grasper
Images in Figure 8-11 show the sequence of a successful biopsy of liver tissue. Frames 1-3 show the positioning of the biopsy jaws over the edge of the liver tissue. Collar actuation is shown in Frames 4-6. Finally, the biopsy grasper was rocked back and forth by moving the wheels. This movement was able to complete the biopsy procedure by tearing the tissue sample from the liver.

8.6 Stapling

The stapling robot was used to place a staple around a branch of the small intestine artery, as shown in Figure 8-12. In this test, the robot was remotely controlled while
driving towards the artery and while maneuvering the stapling arm into position. The actuation mechanism was then used to clamp the staple shut, demonstrating the feasibility of *in vivo* robotic stapling. However, as previously mentioned, the current design of the staple did not work well with grasper design. A more useful staple would be shaped like a ‘V’ so that it could be closed evenly during actuation.

![Figure 8-12: *In Vivo* Stapling of a Mesentery Artery (© IEEE 2009)](image)

### 8.7 Clamping

The next *in vivo* test explored the ability to use the staple robot (without a staple in its end effector) to grasp and manipulate tissue. Early in this test, one of the PMDC motors in the staple robot failed and the robot was removed from the animal. The staple arm was then detached from the body of the robot and reinstalled on the body of another modular robot platform, which had previously been removed from the abdominal cavity. This robot was then reinserted into the abdominal cavity of the animal and used for the remainder of the test. This entire process required less than 5 minutes and serendipitously demonstrated the value of a modular design and the potential for field configurability of these robots.
The reconfigured staple robot was then used to grasp, clamp, and manipulate various tissue and organs within the abdominal cavity. Figure 8-13 shows the robot applying clamping pressure to a portion of the liver. The robot was also used to drag and reposition this portion of the liver, demonstrating the ability of this robot design to generate sufficient traction and grasping forces to provide task assistance for organ manipulation procedures. Literature claims that the force necessary to lift one-third of an average human liver is approximately 4 N [51], and this is a common procedure during operations of internal organs under the liver. The images in Figure 8-13 indicate that it is feasible for the robot platform to produce similar forces.

One final *in vivo* demonstration of the clamping power of the modular robot platform involved stopping the bleeding of a blood vessel. Unfortunately, during this *in vivo* test one of the robot wheels malfunctioned. Consequently, some assistance from a surgeon
was necessary to position the blood vessel within the clamping jaws. Images from the laparoscope video are shown in Figure 8-14. The blood vessel was positioned within the clamp and the clamp closed in Frames 1 and 2. Next, a surgeon cut the blood vessel. Initially, there was no extensive bleeding (Frame 4), which indicated that the clamp on the mobile platform worked successfully. As the robot unclamped the blood vessel in Frames 6 and 7, the blood began to pour out of the vessel. Finally, the clamping jaws actuated over a bleeding blood vessel, shown in Frame 8. When the jaws were closed, the bleeding diminished, as seen in Frame 9 of Figure 8-14.

**Figure 8-14:** In Vivo Images of Clamping a Blood Vessel

### 8.8 Camera

The ceiling pan and tilt (CPT) robot was tested in vivo in a porcine model to demonstrate vision assistance for the surgical team. After insertion, the CPT was
magnetically attached to the abdominal wall, as shown in Figure 8-15. The surgical team controlled the orientation of the video image by moving (panning) the external handle and by using the wireless control system to rotate (tilt) the inner housing relative to the outer housing. Illumination was provided solely by the onboard LEDs. The CPT was able to provide visual feedback of the entire abdominal cavity and was particularly useful in following the mobile sensor robot during exploration, as shown in Figure 8-16.

Figure 8-15: CPT Robot Magnetically Attached to Abdominal Wall (© IEEE 2009)

Figure 8-16: View of Sensor Robot from CPT Robot (© IEEE 2009)
8.9 Cooperative Robots

The modular design facilitated rapid development of robots for different surgical tasks without a complete redesign of the robot. Incorporating the use of the rapid prototyping manufacturing process and ultraviolet-cured glue allowed multiple prototypes to be developed simultaneously in an assembly-line style. End effectors, such as the biopsy grasper and stapling grasper, were developed similarly. Consequently, over a dozen prototypes were constructed in the past year, compared to only a few each year in the past. The increased number of prototypes led to more testing, and likewise, more positive test results. Moreover, the modular platform allowed researchers to design, build and test different robots in tandem.

Multiple robot platforms could be inserted into the abdominal cavity simultaneously as seen in Figure 8-17. Moreover, they operated cooperatively or independently because no cumbersome external connections existed. During in vivo tests, at least two robotic platforms with various payloads were inserted into the abdominal cavity. For example, in one test, a CPT robot was inserted and used to monitor the biopsy robot during a liver biopsy. As previously stated, the navigation of a sensing robot was also monitored as it collected physiological feedback (temperature, pressure, relative humidity) during the biopsy.

Although this type of cooperation is at a fundamental level, it becomes feasible that these platforms can work together inside the peritoneal cavity. Another possibility of cooperation would be assistance from a clamping grasper to help position an abdominal organ or other object while another platform performs a tissue biopsy. Furthermore, multiple in vivo robots can be placed inside the abdominal cavity through a single
incision. Likewise, the size of the incision or diameter of the trocar does not limit the number of wheeled wireless devices that can be inserted for surgical operations.

Figure 8-17: Two Robots Controlled Simultaneously Inside the Abdominal Cavity
Chapter 9

9 Conclusions

9.1 Modular Design

A robotic platform was designed to provide in vivo task assistance during laparoscopic surgery. A battery, main control board, and wheel motors made up the modular portion of the robot platform. The remaining space was dedicated to facilitating various payload options. Rapid prototyping manufacturing techniques allowed the robot prototypes to be produced quickly during research and development. Wheels using a design developed during previous research provided forward, backward and turning motions. Finally, wireless communication with the robot removed any cumbersome wires that caused entanglements in earlier designs.

9.2 Redesigned Biopsy Grasper

Careful attention was paid to the redesign of a biopsy grasper. The biopsy grasper was implemented into the payload area of the modular robotic platform. While the lead
screw actuation mechanism was similar to previous work, the grasper was dramatically altered.

One major area of change was the implementation of overlapping blades. Traditional laparoscopic biopsy graspers have matched jaws that are both actuated with a Teflon-coated cable. Consequently, a “grasp and tear” approach is used to remove the tissue sample. However, the miniature robotic platform could not produce the forces necessary for this type of procedure. The implementation of overlapping blades allowed the robot platform to apply more stress on the tissue, and as a result, tissue samples were severed with less applied force than the “grasp and tear” approach.

Another change was the method of actuation. Instead of a lead screw displacing a Teflon-coated wire, the lead screw was used to slide a collar over the top and bottom jaws, closing them together. The bottom jaw was stationary during actuation, and it was constructed of stainless steel tubing. Meanwhile, the top jaw was constructed on a Nitinol ribbon. The super-elastic ribbon would flex during actuation, causing the top blade to overlap the bottom blade until it came into contact with the bottom jaw.

Lab experiments concluded that approximately 2.83 N was required to slide the collar and close the biopsy grasper. These experiments were performed on an early prototype of the biopsy grasper, which used a thick Nitinol ribbon for the top jaw. For these experiments, minimal effort was placed on optimizing the mechanical efficiency of the grasper. The results were meant to serve as “worst-case” scenario of what would be required to actuate the collar. Also, lab experiments concluded that the lead screw and motor configuration was capable of supplying between 7.3 and 13.2 N, which were much greater than the required forces measured experimentally.
Finite element analysis of the biopsy grasper was performed using the commercial software Abaqus. Results from the laboratory experiments were used to validate the model. For example, the Nitinol ribbon thickness and applied boundary conditions were adjusted until the model best matched the experimental data. Next, simulations concluded that a thin Nitinol ribbon could produce the same jaw force as a thick ribbon. Consequently, a thin ribbon was used in the biopsy grasper design because it required less actuation force without sacrificing jaw force. Finally, simulation results suggested that a top jaw profile angle of 23° was the most feasible option in terms of the actuation force needed, jaw force produced, and the size of the opened jaws.

9.3 Payload Variations

Several different payload prototypes were also designed and developed for the modular robot platform. For example, a sensor package provided wireless feedback of local pressure, temperature, and relative humidity levels. Other physiological data such as acidity levels (pH) could also be implemented into sensor packages.

Various end-effectors were developed for the same type of collar actuation used in the biopsy grasper design. Laparoscopic staples, organ manipulation, and the clamping of blood vessels were all different tools that could be used with the collar actuation. For example, a modular robot initially equipped with a biopsy grasper could be converted in a matter of seconds to a modular robot with a clamping end-effector. The modular design of the end-effectors allowed them to be replaced quickly and easily.

The motor lead screw was also used to actuate a cautery payload. However, the linkage used for the other end-effectors was replaced with a curved Nitinol ribbon. The use of the ribbon allowed the cautery tip to fully retract into the modular robot housing.
This was necessary to prevent possible damage to the cautery tip during insertion, navigation, and retraction. The main battery of the robot platform was not capable of adequately powering the cautery tip for more than a few seconds. Therefore, a battery pack was designed and tethered to the modular robot platform to provide power to the cautery. The battery pack would be placed inside the abdominal cavity ensuring an in vivo cautery tool. A prototype was used to successfully dissect rubber bands in laboratory experiments. The robot moved towards the rubber band, extended the cautery, activated the cautery to dissect the rubber band, and finally, retracted the cautery arm.

Increasing miniaturization of camera technology makes it possible to implement visual feedback in addition to many of these payload variations. For example, a small camera could be added to the sensor package. However, a specific ceiling pan/tilt (CPT) payload variation was developed to replace the visual feedback of the laparoscope during abdominal surgery. An outer housing was designed to fit over the modular robot platform. The housing contained embedded magnets which would allow panning control by a magnetic handle on the exterior of the abdomen. Tilt control was possible by using a wheel motor to rotate the modular robot platform relative to the outer housing. A camera, image processor, and a set of LEDs were placed inside the payload area to provide visual feedback.

9.4 Elastic Tissue Model Results

Abaqus finite element software was also used to analyze the biopsy grasper’s interaction with a soft biological tissue model. Liver tissue material properties were acquired from published literature, and they were used to construct an elastic tissue model. Realistically, soft biological tissue is not elastic. This is especially true for the
large deformations present during a tissue biopsy. However, an elastic model provided a fundamental medium of interaction between the biopsy grasper and the tissue. Effectiveness of the biopsy grasper was evaluated based on the stress that the grasper applied to the tissue model.

Many different simulations were used to investigate a wide range of design parameters. First, results concluded that overlapping biopsy jaws were able to apply 0.2061 MPa to the tissue model, compared to the 0.0517 MPa applied by matching jaws. Next, the clearance, or distance between the top and bottom blades when closed, was investigated. The amount of clearance was controlled by adjusting the bottom blade diameter, top blade diameter, and the position of the top blade with respect to the bottom blade. Results indicated that a small clearance of approximately 0.07 mm applied the most stress to the tissue model.

Two other design parameters proved to have a minimal effect on the elastic tissue model. The top blade was shaped like a semi-circle so that it overlapped the bottom blade. Simulation results indicated that the amount of circle used in the top blade did not have a significant effect on the biopsy grasper performance. Also, the angle of the top blade with respect to the bottom blade did not have a significant effect on the biopsy grasper performance.

The soft elastic tissue model showed no significant changes in the calculated stress as the top blade angle was varied between -10° and 10° with respect to the bottom jaw. However, there were some visible changes in the displacement and distortion of the tissue model. Ideally, the top blade would come into flush contact with the bottom jaw when the grasper is closed.
9.5 Hyper-Elastic Tissue Model Results

Biopsy grasper design parameters were analyzed again after replacing the elastic tissue model with a hyper-elastic tissue model. Deformation of the incompressible hyper-elastic model was much more complex. Therefore, the displacement boundary conditions applied to the blades were reduced in order to avoid convergence problems.

The hyper-elastic tissue model was a more accurate representation of a living soft tissue. For example, measurements were taken from porcine liver, and used to create the reduced polynomial hyper-elastic model. Also, the increased stiffness of living tissue and membranes were better represented by a model that allows modulus to increase as strain increased. The nonlinearity of soft tissue modulus has been well documented in literature.

Once again, overlapped biopsy jaws applied much more stress than matched jaws. Also, the smallest clearance between the top and bottom blades produced the most stress on the tissue model. However, changes in the amount of clearance were much more significant in the hyper-elastic simulations. Elastic simulation results remained somewhat unchanged compared to the results in the hyper-elastic model. Also, the hyper-elastic results indicated that the top blade diameter had more effect on the applied stress, when compared to the effect of the bottom blade diameter. This was because the top blade was displaced more during actuation.

Finally, the angle of the top blade with respect to the bottom blade had a much more noticeable effect on the applied stress in the hyper-elastic simulations. Results concluded that a flush fit between the top blade and bottom jaw was ideal. However, simulation results also indicated that the stress applied to the hyper-elastic model decreased
significantly if the top blade was angled so that its back edges initiated the contact with
the tissue model.

9.6 In Vivo Results

Prototypes of the modular robotic platform equipped with various payloads completed in vivo tests using a porcine animal model. Even though the robot platform was designed with the intention of being a single-use, disposable device, some of the robotic components, such as tools, could be sterilized and re-used. All of the prototypes were able to maneuver and navigate within the abdominal cavity. Furthermore, the modular robots were controlled wirelessly.

Several different payloads completed successful in vivo tests. For example, the sensor package was able to transmit physiological readings wirelessly to a receiver near the surgical table. Also, a hepatic tissue sample was removed from the liver. The laparoscopic stapling payload was tested, but results suggested that a redesign of the shape of the staple would be more successful with this robotic platform.

The clamping payload device easily manipulated the liver. Moreover, the same device was used to successfully stop bleeding from a blood vessel. Finally, a demonstration of cooperative robots was successful when the CPT payload was used to provide visual feedback as other modular robots performed surgical tasks. In fact, many of the in vivo tests involved two or three robots being operated simultaneously.

9.7 Future Work

The immediate future work involves additional in vivo tests to evaluate prototypes. Specifically, additional testing is needed involving surgical task assistance with cooperative robots. For example, modular robots equipped with a clamping payload and
a cautery payload could work together in order to perform a stretch and dissect operation of a blood vessel or possibly a gall bladder removal. Also, a significant amount of work can be done to the tissue models presented in this thesis. Mesh refining can produce a better approximation of interactions between tools and biological tissue.

The modular design of the robotic platform invites the continual development of new payload variations and combinations. For example, a drug delivery robot could be used to coagulate blood in injuries. Also, payload variations could be combined by adding a camera system to current payloads such as the biopsy grasper, cautery, or sensor package. One can imagine a tool belt of different payloads worn by first responders. Once the type of surgical assistance needed is identified, that payload is snapped onto the modular half of the robot and it is inserted into the patient.

These miniature robotic devices make single incision surgery a possibility with robotic assistance. The size and cost of these robots make them economically more feasible to be used in remote areas such as battlefields and even space. Wireless control allows them to be teleoperated remotely by a surgeon before injury trauma increases during transport. Finally, it is important to understand that these devices are not designed with the intent of replacing surgeons. They are meant to extend the hands of skilled surgeons to patients in areas otherwise unreachable with conventional laparoscopic tools and techniques.
References


June: 122-130.
Vivo Soft Tissue Damage Assessment for Applications in Surgery,” *Medical
Engineering & Physics* 32: 437-443.
York: Springer-Verlag.
Layers with Distributed Collagen Fiber Orientations,” *Journal of the Royal Society
Interface* 3: 15-35.
[56] Prange M.T., Margulies S.S., 2002, “Regional, Directional, and Age Dependent Properties
of the Brain Undergoing Large Deformation,” *ASME Journal of Biomechanical
Engineering* 124: 244-252.
Experimental Tensile Behavior of Brain Tissue,” *Biomechanics & Modeling in
Mechanobiology* 5: 53-61.
33*: 367-373.
[60] Chui C., Kobayashi E., Chen X., Hisada T., Sakuma I., 2004, “Combined Compression and
Elongation Experiments and Non-Linear Modeling of Liver Tissue for Surgical
Methodology Using the Finite Element Method and the Extended Kalman Filter to
Identify the Material Parameters of an Organ Model,” *IEEE Engineering in Medicine
and Biological Society*, August: 469-474.
Minimally Invasive Measurement and Characterization of Soft Tissue Response,”
*Medical Image Analysis* 11: 361-373.
*Handbook of Numerical Analysis: Computational Models for the Human Body*,
Elsevier, New Holland, 361-373.
for Linear Versus Nonlinear Elastic Models,” *Joint EuroHaptics Conference &
Symposium on Haptic Interfaces for Virtual Environmental & Teleoperator Systems*.
Organ Mechanical Impedance*, PhD Dissertation.
Characterization of the Liver Capsule and Parenchyma,” *Lecture Notes in Computer
Science* 4072: 150-158.


